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Ultrasonic apnea monitor

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Ultrasonic apnea monitor

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by

Mohammad Bagher Borozjani

A Thesis Submitted to the
Graduate Faculty in Partial Fulfillment of the
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Major: Biomedical Engineering

Signatures have been redacted for privacy

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Ames, Iowa

1980

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INTRODUCTION

Apnea monitors are devices most commonly used to detect life threatening episodes in both adults and infants with respiratory problems. The physician's decision for such patients heavily depends on the data he gets from the monitors, so accurate and reliable information is expected from the apnea monitors.

In the case of infants, apnea monitors are primarily used for detection of apnoeic attack, the absence of respiration for a period of time. Obviously, arterial oxygen tension (PaO_2) monitoring would be of great importance because the results are direct, accurate, and reliable. But, since incision and catheterization are involved, other threats to the life of the infant, like infection and hemorrhage, may appear. It is logical, therefore, to look for other techniques which provide data reliable as PaO_2 monitoring, and yet which require no invasion of the body.

Plethysmographs and monitor pads are now commonly used in hospitals. They detect changes in the volume of the chest or the changes in the resistance of the tissue between a pair of electrodes placed on the chest. These techniques lack the invasive disadvantages of PaO_2 monitoring, but they have their own disadvantages, such as electrode interference, skin irritation, and fall in PaO_2 due to handling of the infant for routine care.

Infants' PaO_2 levels usually fall considerably when they are disturbed. These PaO_2 decreases are observed even during routine handling of the baby such as changing the under-sheet in the incubator, feeding the baby or attaching an electrode to the baby for observation of respiratory or heart rate. Sometimes these decreases in PaO_2 necessitate respiratory support.

In order not to disturb the infant, contactless apnea monitors, such as air-filled mattresses or capacitance change detectors were devised. These monitors and other similar ones detect movements due to respiration rather than respiration itself. Thus, in cases where the upper airway is blocked, according to Hering-Breuer Reflex theory, the infant would make an effort for respiration even though there is not true respiration. These cases are not detected by contactless monitors. Furthermore, some of these techniques like air-filled mattresses, can be used only for a short time period; and some others such as capacitance-change detectors are particularly sensitive to moving objects around the baby also.

In this thesis, a novel contactless technique for monitoring apnea is described which lacks most of the disadvantages of other contactless monitors, and, with the help of a minicomputer, detects upper airway occlusion.

This technique utilizes a pair of ultrasonic air

pressure transducers functioning as a transmitter and a receiver. The transmitter sends pressure waves (40 KHz) toward chest of the infant, and the second transducer picks up the reflected waves. An electronic circuit connected to the transducers detects the phase shift of the reflected waves with respect to the transmitted waves. When the chest wall is moving, the phase shift changes, and when there is no chest wall movement the phase shift remains constant, a condition which is detected and used to trigger an alarm.

The computer section of this technique is for detection of upper airway occlusion (U.A.O.). The waveform of the chest wall movement during U.A.O. is considerably different from the normal waveform. The computer continuously processes the waveform and turns on the alarm when the specific changes of U.A.O. in the waveform appear.

For primary testing of this monitor, a respiration pump was used. A plate attached to the piston of the pump was used as a reflector. When the piston was stopped, the alarm sounded.

For further evaluation of this technique, animal trials were instituted. The subjects were young mongrel dogs ranging from 5 to 10 lbs. The dogs were anesthetized to prevent body movements. A thermistor was placed in the lumen of an endotracheal tube and then the dog was intubated. Conventional respiration waveforms from a thermistor bridge

were then taken. Though the waveforms from the two methods were not exactly the same, both indicated respiration very well. During these trials, the buzzer of the contactless monitor never sounded, i.e., no false alarm was observed.

To verify the validity of the computer program, the endotracheal tube was clamped (artificial U.A.O.) several times. The computer detected the occlusion in most of the trials.

LITERATURE REVIEW

Apnea and Causes

Apnoeic attack, defined as the absence of the breathing for a period of more than 20 seconds (1), is a major threat to the lives of infants of very short gestational (life in the uterus) age (2). The condition has a high mortality rate and is likely to be a possible cause of hypoxic brain damage among survivors (3).

Apnoeic attacks should not be mistaken with Sudden Infant Death Syndrome (SIDS) which has been established as a distinct disease entity (4, 5). In a significant proportion of SIDS cases viral infection is present. In addition, most cases of SIDS occur between 4 and 16 weeks, suggesting some physiological vulnerability during this age period (6), while apnoeic attack is usually seen during the first days or even during the first hours after birth.

Apnoeic attacks may occur in association with a variety of disorders, but the most convincing explanation is physical immaturity (1). In order to expand the chest and thereby increase the negative intrapleural pressure so that air can enter the lungs, a certain degree of rigidity of the chest wall is necessary and is normally provided by the ribs. Since newborn infants, particularly prematures, do not have a rigid bony thorax, some difficulty in expanding the lungs

occurs (7).

Slow and late appearance of alveoli may be another reason confirming the explanation of immaturity for apnea in infants. A mature newborn has about 20 million alveoli. By the time the individual reaches maturity, the number of alveoli reaches about 300 million; thus, much of the development of the alveoli occurs after birth (7).

Some premature infants simply forget to breathe while they are in deep sleep. These cases are believed to be due to immaturity of autonomic nervous system control mechanism. Thus it may not be far from fact if one accepts the general explanation of immaturity as the cause of apnea in infants.

Techniques for Monitoring

Because of the significant threat of apnoeic attack to the lives of infants, monitoring respiration of newborns, primarily in cases of premature birth, is of considerable importance. The most reliable and accurate method for monitoring respiration is monitoring the PaO_2 level in the arterial blood. In this method a catheter is inserted, usually into an umbilical artery, until the oxygen electrode tip is in the lower aorta. Although this method directly monitors oxygen tension level and hence accurate and reliable results are obtained, incision, asepsis and catheterization are involved; therefore, some side threats to the

patient, like hemorrhage and infection, are created. Because of these threats and the delicate and time-consuming procedure of setting up the system, many physicians and researchers have been searching for noninvasive and safe techniques to monitor respiration.

Several of these noninvasive methods described here utilize thermistors in respiration detection (8).

Thermistors are electrical devices which exhibit an inverse relation between temperature and resistance. They are placed in the oral or nasal airflow. The exhaled gas is at a different temperature compared to the inhaled gas. As a result, the resistance of a thermistor element will change with inspiration and expiration if it is placed in the airflow.

Thermistors are placed in the lumen of an endotracheal tube and then the patient is intubated. The output of the thermistor is fed into an electronic circuit for monitoring and recording. This technique necessitates anesthetization of the patient, an undesirable situation because anesthetization relaxes the control mechanism of the nervous system and causes the patient to make less of an attempt for respiration, thus increasing the chances of apnea.

Pharyngeal probes, using thermistors, have also been developed (8). One which is commonly used, utilizes a 7F (for adults) of 5F (for children) nasogastric tube. The

tube with the thermistor inside is positioned in the posterior pharynx via a nontoxic flexible catheter. The posterior pharynx receives nasal and oral airflow and can be catheterized comfortably and safely. An electronic circuit is also needed to be added to the system to condition the signal and to detect respiration. The disadvantage with this technique is that the pharyngeal placement of the probe is not well tolerated in children. Also, after many hours of continuous use the signal may become attenuated as a result of mucous accumulation.

Other, noninvasive, techniques of respiration monitoring have been developed. These techniques can be divided into two groups, plethysmography and motion detecting techniques.

Plethysmography generally measures change in volume. A body plethysmograph was the first apparatus used for measuring ventilation in the newborn (7). In this technique the infant would be placed in a closed chamber, while the infant's mouth is kept outside the chamber. With respiration, the volume of the chest changes and consequently the pressure inside the chamber changes. The fluctuation in the pressure is picked up by a pressure transducer attached to the chamber.

Using this technique, early workers were bothered with the air leak around the infant's neck. To overcome this

problem Karlberg and Koch (9) used a reverse plethysmograph. In this adaptation the infant breathed through a mask attached to a wide tube connected to a plethysmograph.

Major disadvantages of the masks are that they disturb the infant's normal respiratory pattern and produce an intolerable increase in the dead space. Low dead space, low resistance nasoalves have been constructed (10), but the general problem with such devices as well as with the body plethysmograph still persists; i.e., the system, being a rebreathing one, must be used only for a short period of time to avoid significant CO₂ accumulation.

Simpler plethysmographs such as: i) capacitance plethysmograph, ii) elastic-resistance strain-gage plethysmograph, and iii) impedance plethysmograph, have none of the problems associated with the body plethysmograph and yet are less expensive and easier to handle.

The first of these simpler methods measures the change in capacitance between the skin and an outer rigid electrode. The second method measures the change in the resistance of a conducting liquid in an extendable rubber tube placed around the body. These two methods are usually used to measure blood flow in extremities (11). An increase in volume of blood in a limb results in a decrease in separation

of the plates and an increase in capacitance, or results in an increase in the length of rubber tube of the related plethysmograph. In the first technique the electrical stability is a problem, and in the second method the electrical resistance is very low, and also the relation between resistance of the conductive liquid and the length of the tube is not linear.

A related method, impedance plethysmography, is based on the fact that changes in a tissue volume are accompanied by changes in the electrical resistance of the tissue. The electrical resistance can be measured by attaching electrodes to the tissue and requiring a "low-level" current to flow through the tissue. The voltage seen across the electrodes is proportional to the electrical impedance (11, 12).

The volume of the lungs changes with respiration and therefore the resistance between a pair of electrode attached to an infant's chest will change. This method requires a high frequency oscillator (50-100 KHz) which sends a current (less than 0.3 ma) through electrodes placed on the chest wall of the infant (12).

The general disadvantages of plethysmographs are skin irritation, interference with patient care, and slow application in emergencies (13). Also, there might be false alarms due to lead or electrode failure.

Instead of plethysmographs, monitor pads can be used. Monitor pads, usually made of disposable foam rubber, have four conductive vinyl surfaces, each representing standard electrocardiography (ECG) limb-lead points (RA,LA,RL,LL), each vinyl surface is connected to a snap to be attached to external wires. Monitor pads are worn on the back or around the chest of the infant. They are often used to obtain the ECG, but can also be used as an apnea monitor by measuring the impedance changes between the transthoracic RA-LA leads (13). By experiment (13), it has been shown that monitor pads do not have most of the disadvantages of the plethysmographs described.

Both monitor pads and plethysmographs do share a common disadvantage. They are extremely sensitive. So much so that they usually pick up the heart pulse as well as respiration.

A rather different technique which lacks the above disadvantage is the abdominal respiration detector. Increase in volume of the abdomen on respiration increases the radius of curvature of the anterior abdominal wall (14). The detector consists of a flat cylindrical capsule about 20 mm in diameter and 3 mm deep. The open end may be covered with a thin rubber diaphragm, or it may be applied directly to the abdominal wall to which the capsule is secured with adhesive tape. A nozzle in the side of the capsule is

connected by a length of flexible tubing to a pneumatic transducer. This may be either a sensitive capacitance manometer (14) or an electro-magnetic movement transducer (14) attached to a similar diaphragm capsule.

An increase in the volume of the abdomen causes a decrease in the volume and consequently an increase in the pressure of the cylinder. Volume changes of less than 0.1% can be detected (14). In this technique respiratory movements are reproduced with remarkable faithfulness provided the subject does not change position.

All the techniques described so far regardless of their advantages and disadvantages are accompanied by a potential danger. They all have something in contact with the patient. Observation of infants during routine care has shown handling for any reason, such as X-ray, heart rate, and respiration observations, often causes a fall in the arterial oxygen tension (15). Although the infant usually recovers spontaneously, a series of procedures, one following closely upon the other, may cause a prolonged fall in PaO_2 necessitating respiratory support. In cases of very ill patients even minor disturbances like attaching an electrode to the infant or changing the under sheet in the incubator causes a fall in PaO_2 of as much as 30-40 mm Hg from a normal value of 80-90 mm Hg (15).

To avoid disturbing the infant, contactless devices

for monitoring respiration in infants are desirable. The air mattress, used in some hospitals, is listed under the category of contactless devices for monitoring respiration. A heated thermistor is placed inside an air filled mattress. Breathing movements cause air to flow from one section of the mattress to another, cooling the thermistor and thus providing a measure of respiratory movements.

The major disadvantage of the air mattress for recording respiration patterns is that over the period of a few breaths, the sensitivity changes. A second disadvantage is the double peaking effect, which means that the thermistor becomes cool on both inspiration and expiration, thus providing two signal peaks per respiration cycle.

A different contactless respiration monitor, the under-mattress pressure sensor, utilizes a thin 150 mm square pad in which electrode plates are incorporated. This pad is placed under or slightly above the chest. Respiratory movements change the capacitance between the electrode plates. Comparison of this device with impedance respirometer has shown that the changes in sensitivity as the infant moves about are quite considerable; e.g., the sensitivity changes by a factor of 10-20 for different positions to which the infant moves. In some cases, the sensitivity changes over the interval of a few breaths without any apparent change in

the body position (16).

A capacitance sensor can also be used to monitor the respiration without contact. A pair of electrodes is placed near the body. Respiratory movements change the capacitance between electrodes. The measurement of capacitance changes is achieved by applying a 200 KHz signal across the electrodes and detecting the current flow with a phase sensitive amplifier. The disadvantages of this system are: i) it is too sensitive to people moving nearby, and ii) the incubator should be electrically shielded. This type of system cannot readily be made portable (16).

Another technique which lacks the disadvantages of the two mentioned contactless monitors is the Radar Respiration Monitor. This system senses chest wall movements by detecting changes in a reflected electromagnetic micro-wave signal. A microwave source sends signals in form of radio frequency energy to the chest. The detector output is a periodic wave as the infant's chest moves back and forth. The period of this wave corresponds to the time the chest covers half the microwave wavelength (30 mm if a 10 GHz source is used).

This system has proved to be very satisfactory where electrodes and wires are hazardous. Nevertheless, the 10 GHz microwave frequency of the most readily available miniature microwave components is not ideal for chest wall movement. With very large inspired volumes, as in gasps,

the chest wall movement may be more than half the microwave wavelength, and this results in a distorted signal. Besides, the output from the detector may give a very distorted signal because it depends on the whereabouts on the standing wave pattern as chest movements occur.

All the techniques described in this chapter, except for the direct measurement of PaO_2 and the pharyngeal probe, detect either changes in the volume or movement of the chest. The characteristics of these techniques bring about a problem for monitoring apnea. Upper airway occlusion may not be detected because obstruction of the upper airway in infants may be associated with continued respiratory effects without air entering the lungs (17). These efforts cause changes in volume, electrical resistance, or position of the chest; thus, these changes are detected by monitors as respiration while there has not been true respiration.

Treatment of Apnea

Thus far, no effective treatments for apnoeic attacks have been found, primarily because the causes of apnea are not known. If the explanation of immaturity for apnoeic attacks is correct, then the best treatment is to wait for the infant to grow, providing apnoeic attacks are detected and treated. Nevertheless, there have been some ways

suggested or tested to prevent apnea. One method advocated for treatment of apnea is raising the ambient oxygen concentration, but it is often ineffective and may expose the infant to the risk of hypoxia between attacks (1).

Another way of treatment of apnea is the use of continuous positive airway pressure (C.P.A.P.). Applying C.P.A.P. to infants with severe recurrent apnea appears to provide respiratory drive for the infant, perhaps due to stimulation of the Hering-Breuer inflation reflex. This technique has proved to be effective in decreasing the rate of apnoeic attacks, but can not remove the attacks totally. Thus, the use of apnea monitors which provide reliable data and expose the infant to minimum side effects of monitoring is of great importance.

NEW TECHNIQUE AND DESIGN

The literature review indicates that monitoring of respiration is of great importance. However, we also learned that some of those techniques pose threats or are of low reliability. Among those techniques, contactless methods were found to be least harmful.

The contactless technique described here eliminates both the cost and accompanying skin problems of electrodes. In addition, with the help of a computer, upper airway occlusion (U.A.O.), an important concern of physicians, can be detected.

Basic Criterion

This ultrasonic technique is based on the standing wave phenomenon. A low level ultrasonic (40 KHz) air-pressure transducer¹, placed in front of the infant's chest or abdomen wall, sends pressure waves to the infant. The waves are reflected by the infant's chest wall with the same frequency, but with a different phase angle. A transducer similar to the transmitter, picks up the reflected waves and produces 40 KHz AC voltage at its output. If the transducers are side by side (see Figure 1), the phase angle of the reflected wave at the pick-up transducer is

¹Model 70100 40 KHz, Linden Laboratories, Inc., Box 920, State College, PA, 16801 (814-355-5491).

$$\phi = 2\pi \cdot 2 \cdot l \cdot \frac{F}{V} + K \quad (1)$$

where

l = distance between electrodes and the chest wall

F = frequency of oscillation

V = velocity of sound in air = 330 m/sec

K = constant

Since V , F , K are constant, changes in ϕ are proportional to changes in l ,

$$d\phi = \frac{4\pi F}{V} \cdot dl \quad (2)$$

According to Equation (2), if l is constant, then there is no change in the phase angle. In other words, continuous change in ϕ indicates continuous change in l , the distance between the electrodes and the infant, and thus indicates that the chest wall is moving.

The change in the phase angle difference between the transmitted and the reflected waves is proportional to the displacement of the infant's chest wall. In order to find the proportionality between the displacement and the phase angle difference, we must go back to Equation (1).

$$\phi = \frac{4\pi F l}{V} + K$$

$$\frac{F}{V} = \frac{1}{\lambda} \quad (\lambda = \text{wavelength of the waves}) \quad (3)$$

$$\phi = \frac{4\pi l}{\lambda} + K$$

The maximum change in phase angle difference can be 2π radians; thus

$$\Delta\phi = 2\pi$$

$$2\pi = \frac{4\pi}{\lambda} \cdot \Delta l$$

$$\Delta l = \frac{2\pi \cdot \lambda}{4\pi}$$

$$\Delta l = \frac{\lambda}{2} \quad (4)$$

on the other hand,

$$\lambda = \frac{V}{F} = \frac{330 \text{ m/sec}}{40 \text{ KHz}}$$

$$\lambda = \frac{330 \times 10^3 \text{ m/sec}}{40 \times 10^3 \text{ rad/sec}} \times 10^3 \frac{\text{mm}}{\text{m}} = 8.25 \text{ mm}$$

$$\frac{\lambda}{2} = 4.12 \text{ mm}$$

Thus for every 4.12 mm of chest wall displacement, there would be 2π radian change in the phase angle.

Circuit Design

In order to find the phase angle difference between the transmitted and the reflected waves, one should know a little about the signals going in and out of the transducers. The input and output signals from the transducers are sinusoids with different amplitudes; but with the same frequency of 40 KHz. The amplitude of the signal from the pickup transducer, however, varies with changes in the

distance between the infant's chest wall and the transducers.

These fluctuations in the amplitude cause some problem in measuring the phase angle difference between the two waves, because in the technique we have chosen for measurement of phase angle difference, the amplitudes of the waves should be constant. To better understand how the measurement of phase angle difference is made, consider Figure 2. Figure 2 (a and b) shows a pair of square waves with the same amplitude, but with a lag time between the two waves. Figure 2 (part c) shows the summation of the two waves. If this waveform is rectified, a series of pulses with equal widths will remain. The widths of all pulses are equal and proportional to the phase angle difference. As the phase angle difference increases, the width of the pulses will decrease and vice versa.

A block diagram of the apnea monitor is shown in Figure 3. In Figure 3, block 1 is a sine wave generator. The oscillator drives the transmitter transducer, and the reflected wave drives the receiver transducer. The output of the receiver transducer is a 20 mV peak-to-peak sinusoid. To reduce the noise interference with this low amplitude signal a 10 K Ω resistance is placed across the output of the receiver transducer. The output is then fed into 2 stages of linear amplification in order to have a signal comparable in amplitude to the output of the

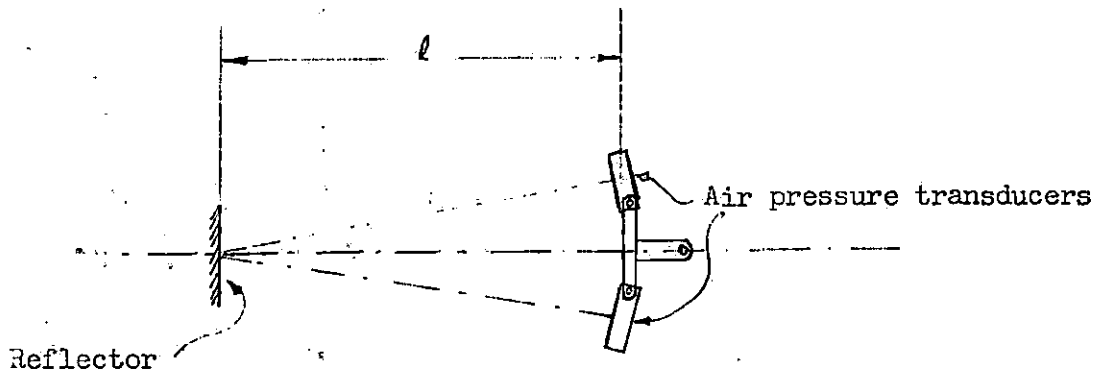


Figure 1. Illustration of relative positioning of the transducers and the chest

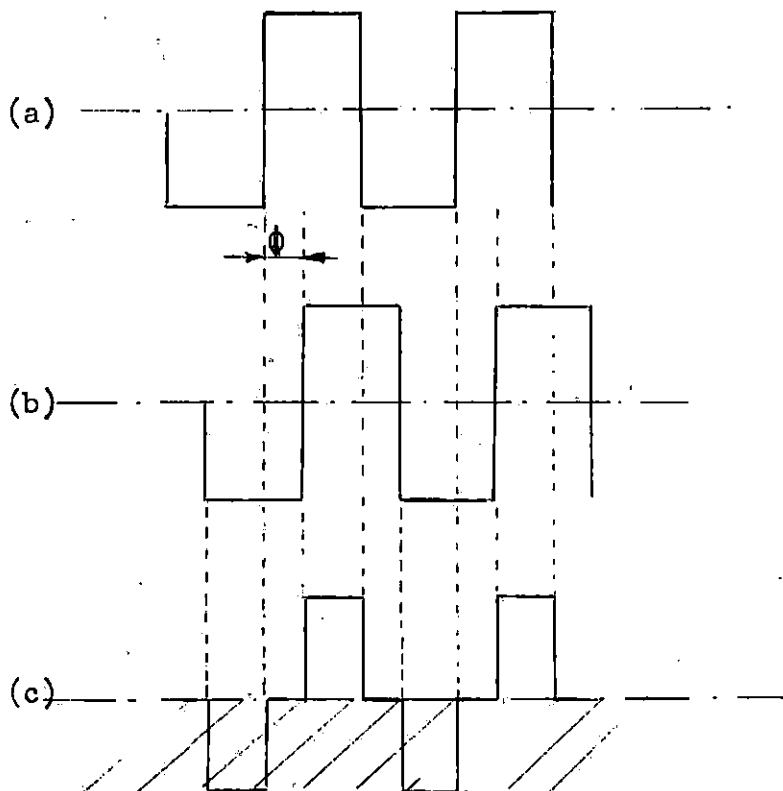


Figure 2. A geometrical demonstration of why the summation of two equal amplitude square waves represent the time lag between the two waves. ϕ is the time lag between (a) and (b). Part (c) is the sum of (a) and (b)

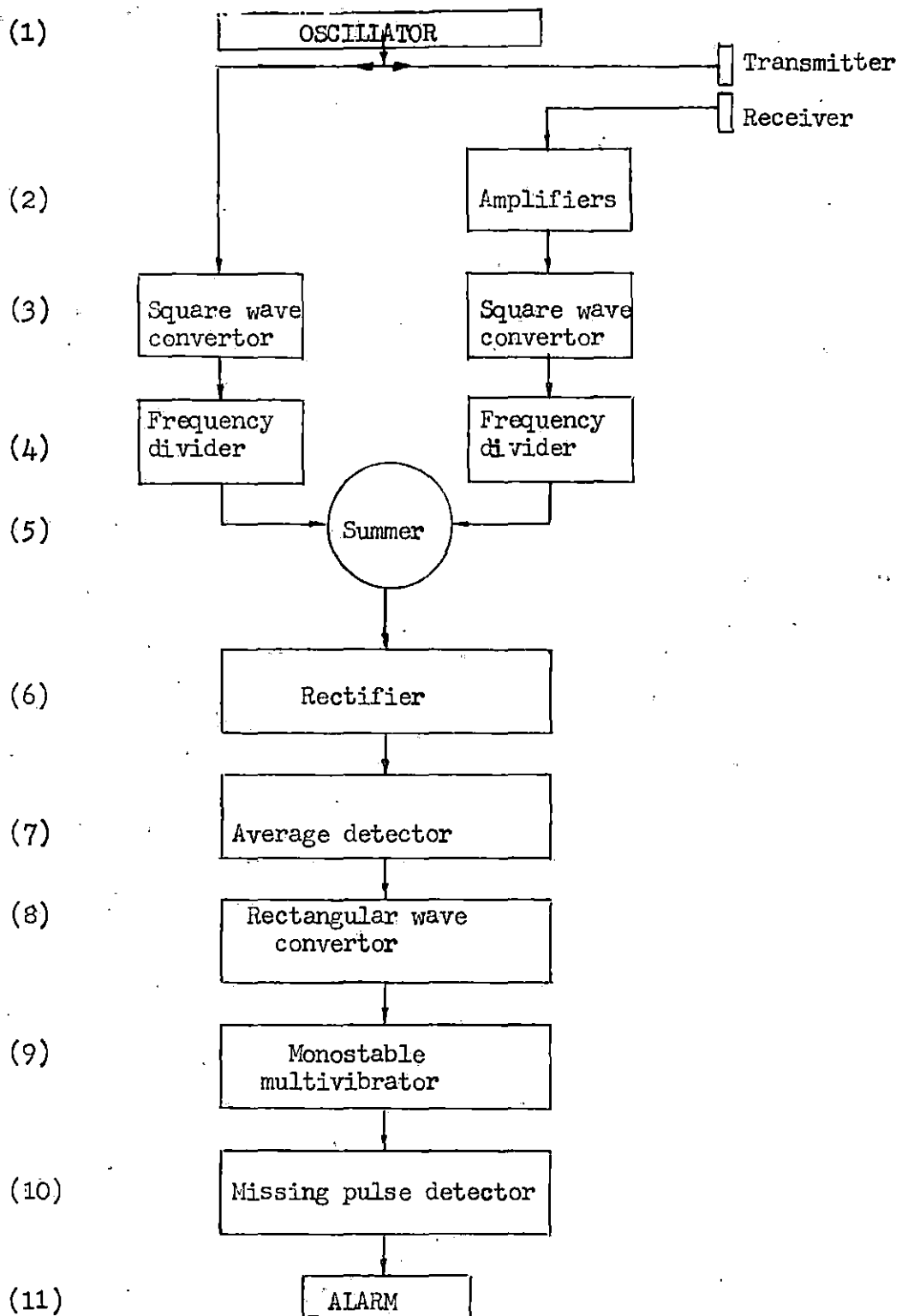


Figure 3. Block diagram of the ultrasonic apnea monitor

oscillator. Two stages of amplification rather than one are used to avoid reduction of the bandwidth.

Block 3 in the diagram produces a pair of square waves at the same frequency of the oscillator. One of these square waves is in phase with the transmitted signal and the other square wave is in phase with the received signal.

In block 4 two simple monostable multivibrators halve the frequencies of the square waves and increase the apparent wavelengths to overcome the double peaking form of the respiration pattern due to large displacement of the chest wall in deep respiration such as gasps or sighs. In deep respiration the displacement of the chest wall may be larger than half of the wavelength (4 mm). In such cases, the phase angle will repeat itself and causes the respiration pattern to have two peaks for each cycle of breathing. This is considered a distorted signal and should be eliminated.

At block 5 a CMOS fast operational amplifier is used to add the outputs of the monostable multivibrators of block 4.

Blocks 6 and 7 are simply a half-wave rectifier and an average detector, respectively. The output of the average detector is a low-frequency voltage proportional to the phase angle difference between sinusoids at the transducers. The magnitude of this voltage fluctuates with respiration.

The waveform of this fluctuation is the reproduction of the chest wall movement, and is an alternating voltage added to a constant voltage.

Blocks 8 through 11 are the alarm section. Block 8 is a comparator which converts the fluctuating voltage, proportional to the phase angle difference, to a rectangular waveform. At the negative going edge of each of the rectangular cycles, a pulse is generated by a monostable multivibrator. Each pulse represents a complete cycle of respiration.

Block 10 is a missing pulse detector. This block triggers the alarm whenever no pulse is generated during a specified period of time.

Detailed circuitry and the waveforms of signals at several points along the path are given in Appendix A.

Computer Programming for Detection of Upper Airway Occlusion

The computer portion of this monitor, designed to detect Upper Airway Occlusion (U.A.O.), is based on the following criterion. During U.A.O. the respiratory waveform is both decreased in rate and prolonged in width. Experiments on puppies used for evaluation of this technique have showed that the respiration rate decreases to two-

thirds of the original rate, and the duration or width of the waveform increases by a factor of 1.5. These characteristics are used to detect U.A.O. A generalized flow diagram of the computer program is in Figure 4.

Block 1 of the flow chart in Figure 4 measures the width and the period of the respiratory cycles and stores them in designated memory locations. This is accomplished by repeatedly taking samples and comparing them with a threshold. Whenever the sample is greater than the threshold, it indicates the onset of a respiration cycle. The computer then waits for 10 milliseconds, increments the time count index, and then takes another sample. This loop is repeated until a sample is smaller than the threshold. The time count index represents the width of the cycle. The action of taking a sample and waiting for 10 milliseconds continues until a new cycle is detected. Now the time count index indicates the period of the cycle. The whole procedure will be repeated six times.

The second block simply finds the average values of the widths and the periods of the first five cycles.

In the third block, the sixth cycle is compared to the results of block 2. At this time only the widths of the cycles are compared, because changes in the width exceed other changes in the waveform of the cycle during

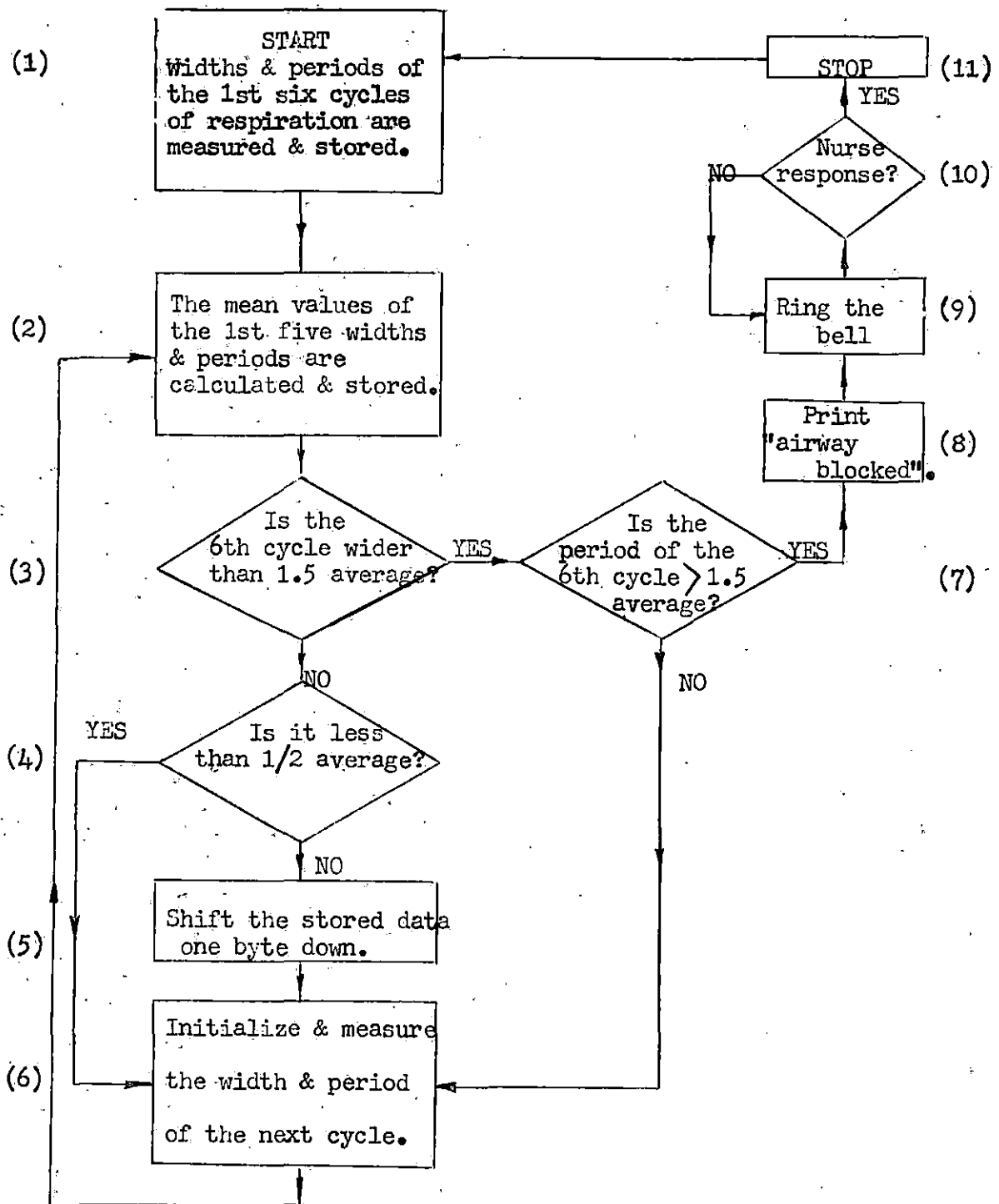


Figure 4. Generalized flow chart of the U.A.O. detecting program.

U.A.O. If the width of the 6th cycle is greater than 1.5 times the average value, there might be an U.A.O. occurring. If not, then the computer checks if there has been any body movement. This is done by block 4. Experiments on dogs have shown that when the body is moving, the widths of the cycles are very small, often less than $1/3$ of the mean value. In block 4 the width of the sixth cycle is compared to $1/2$ of the average value. If the width is more than $1/2$ the average value, the cycle is considered normal and the computer proceeds to the fifth block. If the width is less than $1/2$ the average value, block 5 is by-passed and the flow of the execution of the program shifts to block 6.

In block 5 the data regarding the first cycle are erased and the next five cycles are considered as the cycles for calculation of the average values of width and periods.

Block 6 initializes the program and measures the width and period of the next coming cycle, and then shifts the flow of the program to block 2.

The computer will go through this loop from block 6 to block 2 so long as respiration is normal. If the respiration becomes abnormal then the computer branches out of the loop and goes to block 7.

In block 7 the period of the sixth cycle is compared to

1.5 times the average period. If the period has not increased, that cycle is considered as a deep breath or sigh and ignored. If the period is more than 1.5 times the average value, then an U.A.O. has occurred. Block 8 prints the message "Airway blocked" on the screen of the terminal.

Blocks 9 and 10 ring the bell, until the attendant responds and stops execution of the program. The detailed program is in Appendix B. A digital PDP 8/e minicomputer was used.¹

¹Digital Equipment Corporation, Maynard, Mass.

EXPERIMENTAL EVALUATION

Method

In this section evaluation of the general operation as well as the reliability and consistency of the ultrasonic respiration monitor are described. The general operation was tested by checking the detection of motion, and the reliability was verified by comparing the outputs of this device and that of a reliable monitor.

In the first stage of this experiment the general operation of the system was tested. For this purpose a plate was attached to a rod secured to the piston of a respiration pump¹ as shown graphically in Figure 5. When the pump was turned on, the piston would slide back and forth; thus, the movement of this plate would be a simulation of a chest wall respiration pattern. The apnea monitor should have turned the alarm on whenever the pump was off for a predetermined length of time.

In the second stage of this experiment, using a live subject, the consistency and similarity between the waveforms of the ultrasonic monitor and another reliable monitor were verified. As a live subject, a mongrel young dog was chosen for two reasons:

- 1) there was no safe and easy access to human infants,
- 2) a puppy dog would breathe shallower than a grown dog, and hence there would be more similarity between this case and the case of a newborn.

¹Model #507, respiration pump, Harvard Apparatus, Hillis, Mass., 02054.

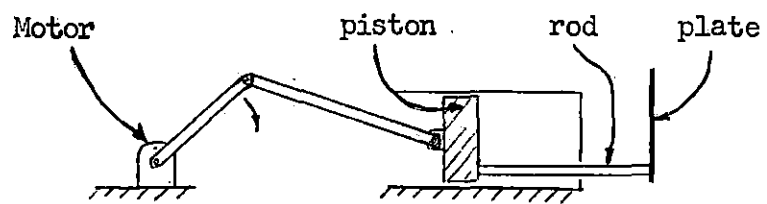


Figure 5. A plate attached to the piston with a rod to simulate the chest wall respiration pattern

For evaluation and verification of the accuracy of the ultrasonic monitor, a thermistor was used. As mentioned earlier in literature review, thermistors detect changes in the temperature of the air surrounding the device. When a thermistor is placed in the oral or nasal airflow, it will detect the temperature changes during expiration and inspiration; therefore, it reliably detects respiration. Furthermore, if the airway is blocked, since there is no air flow around the thermistor element, the blockage is reflected in the temperature of the thermistor by a constant decrease, a case which very well indicates upper airway blockage.

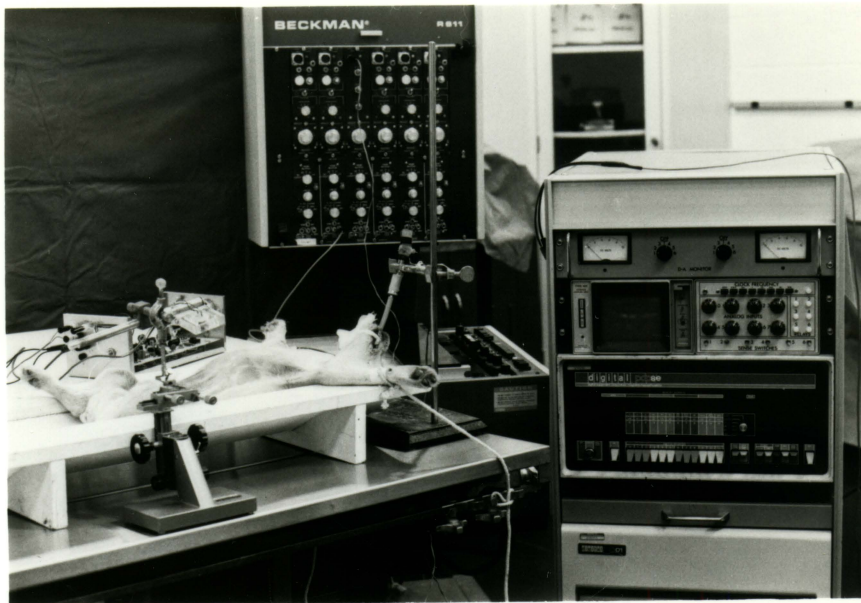
In this experiment, a thermistor was placed in the lumen of an endotracheal tube and the dog was anesthetized with pentobarbital (0.25 ml/kg) and then intubated. Anesthetization makes the dog more cooperative during intubation and prevents the dog from changing position during the experiment. Fortunately, this situation is similar to that of a premature infant which rarely moves about.

After intubation, the dog was shaved on the chest and abdomen and placed on a V-board in the supine position. The thermistor output was fed into the thermistor coupler of a Beckmann recorder. Then the transducers were placed in front of the abdomen of the dog and the distance between the

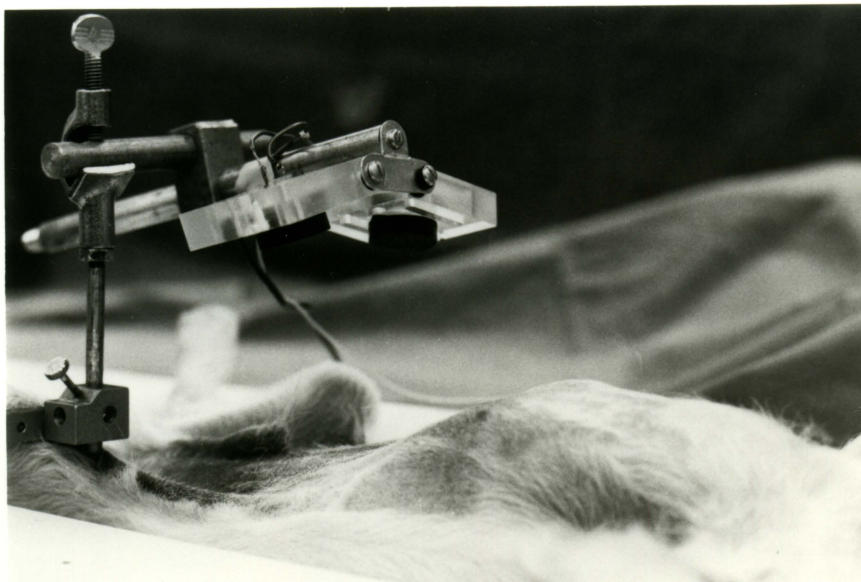
transducers and the abdomen was adjusted in such a way that normal lines protruding from the transducers intersected at a point near the skin surface (see Figure 6). The transducers were then connected to the monitor and the output of the monitor was fed into both the minicomputer and the Beckmann recorder.

First, normal patterns of respiration were recorded. Then the dog was moved to different positions and recordings were obtained. Finally, using a hemostat, the endotracheal tube was clamped while the waveform pattern was being recorded. An assistant marked the recordings whenever upper-airway occlusion was applied or removed, and whenever occlusion was detected by either the computer or the monitor.

Figure 6. Illustration of the set-up of the experiments:
(a) the anesthetized dog is intubated and placed in supine position on a V-board. The transducers are placed in front of the abdomen and connected to the ultrasonic apnea monitor. The output of the monitor is fed into both the Beckman recorder and the minicomputer, (b) the transducers are placed in front of the area immediately below the thorax in such a way that the normal lines protruding from the transducers intersect near the skin surface



(a)



(b)

RESULTS

The experiment was run several times. In the first experiment, recordings of normal respiration patterns were made. Figure 7(a) shows a sample of that recording. Figures 7(b) and 7(c) show the respiration pattern while upper airway occlusion was applied at the end of inspiration, and expiration, respectively. Note that there are two peaks in the monitor trace for every cycle of respiration.

In the second experiment, a pair of frequency dividers were added to the monitor to eliminate the double peaking effect. The recordings of respiration while the dog was in different positions were then taken. Figure 8(a) is the recording taken while the dog was lying on its side. Figures 8(b) and 8(c) show the recordings while the dog was in the supine position and the transducers were facing the dog's chest and abdomen, respectively. Note that in Figures 8(b) and 8(c), heart beats are picked up and recorded by the monitor.

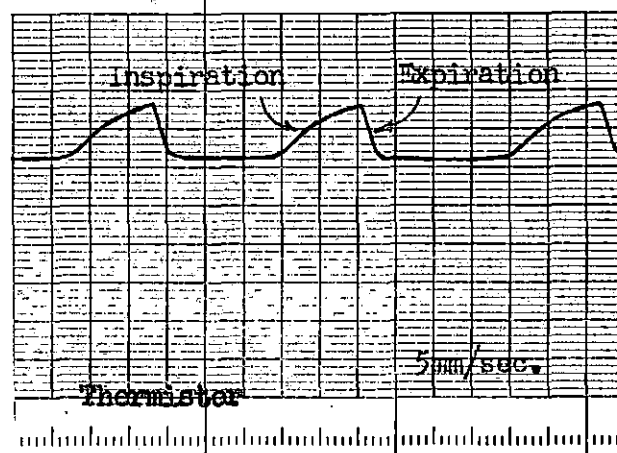
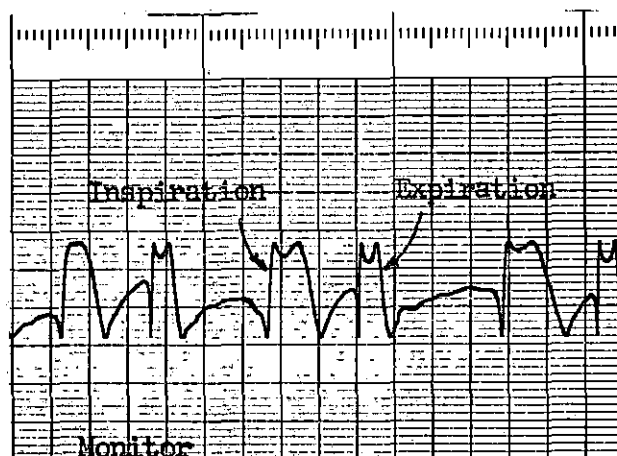
During different phases of respiration (in the middle or at the end of inspiration or expiration) the upper airway was occluded and the waveforms were recorded. Figure 9 shows these recordings. In all cases except when U.A.O. was applied at the end of expiration, the monitor alarm was

Figure 7. Recordings of respiration pattern:

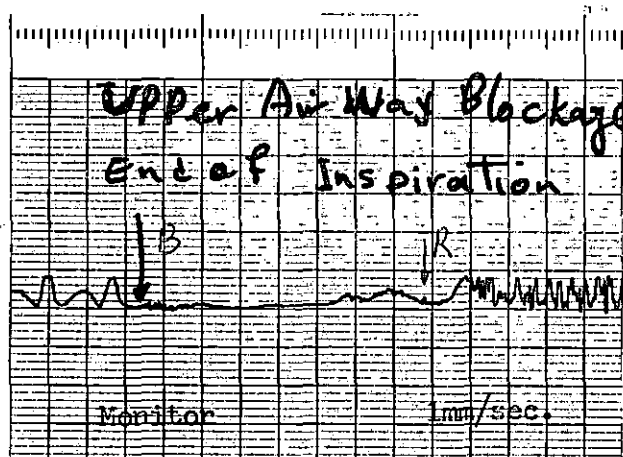
- (a) normal respiration pattern,
- (b) respiration pattern with U.A.O. applied
at the end of inspiration,
- (c) respiration pattern with U.A.O. applied at
the end of expiration

(B indicates application of occlusion and R
indicates its removal)

(a)



(b)



(c)

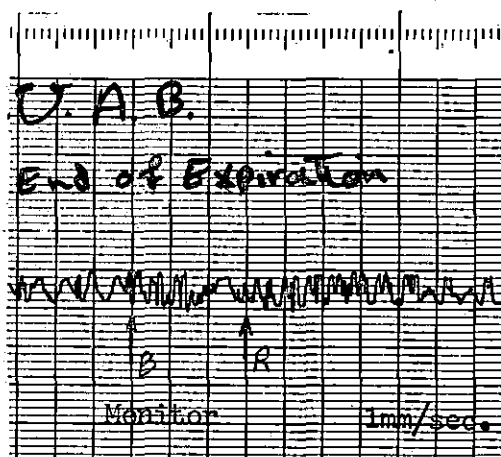
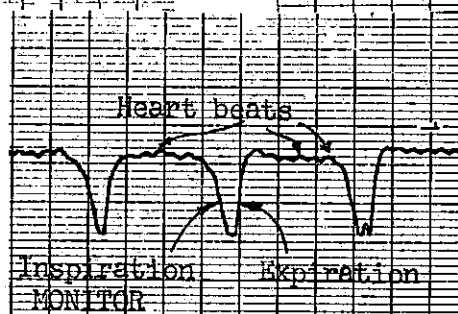
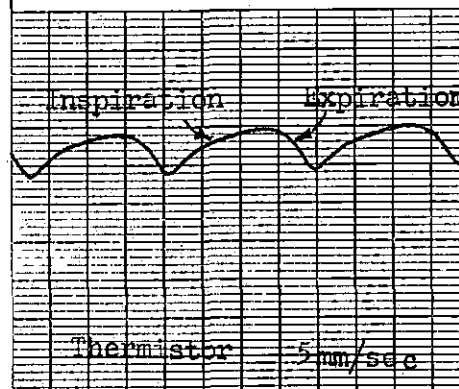
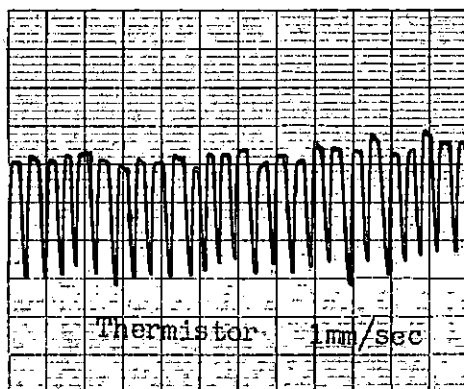
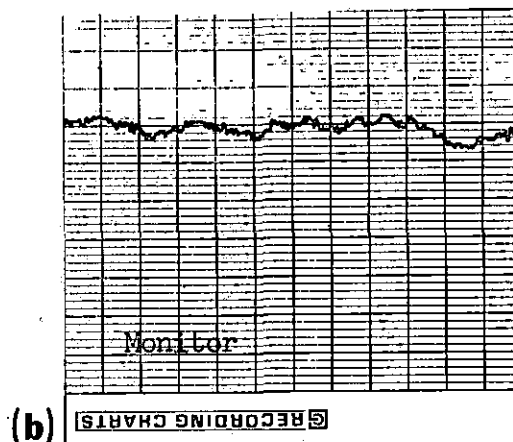
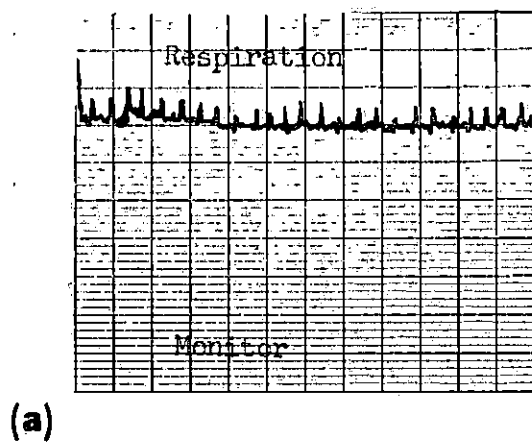


Figure 8. Recordings of respiration with the dog in different positions: (a) the dog laid on its side, (b) the dog in supine position and the transducers facing the chest, (c) the dog in supine position and the transducers facing the abdomen



(c)

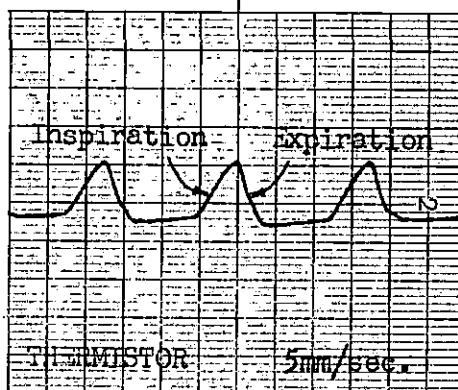
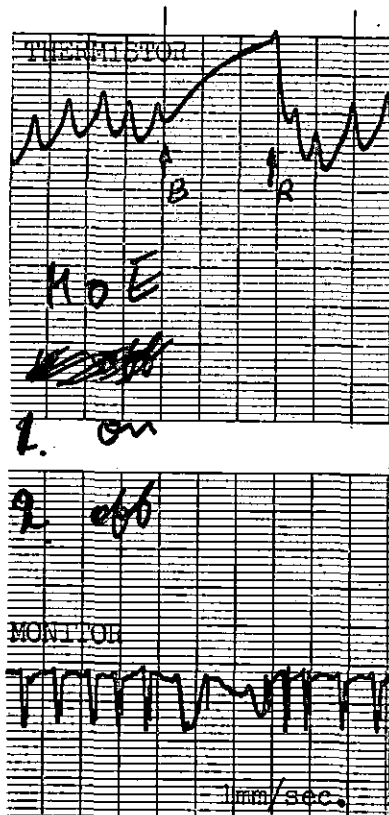
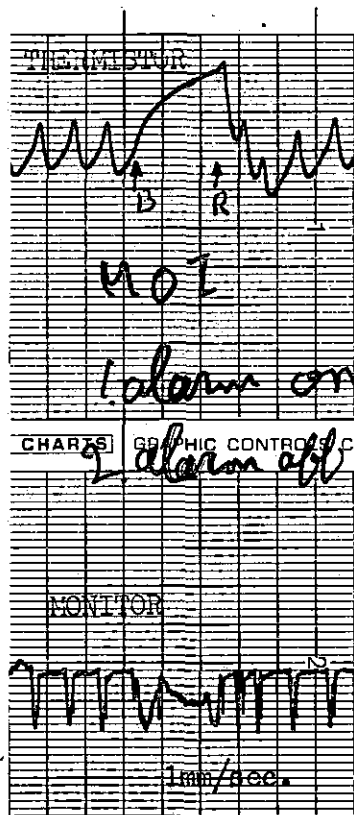


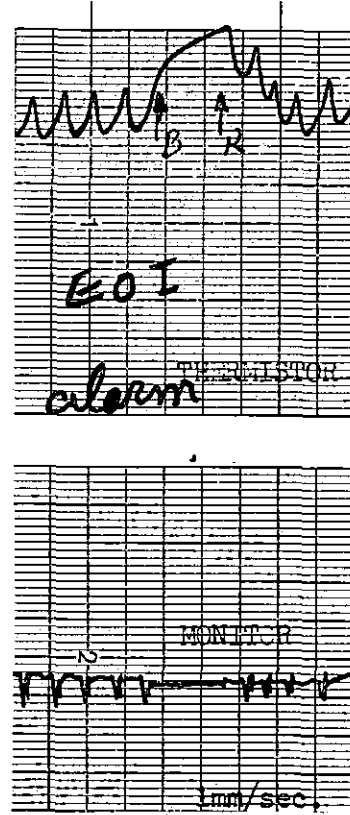
Figure 9. Changes in the pattern of respiration when U.A.O. is applied at the (a) middle of expiration, (b) middle of inspiration, (c) end of inspiration, (d) end of expiration. Note that only in case (d) U.A.O. was not detected by the monitor



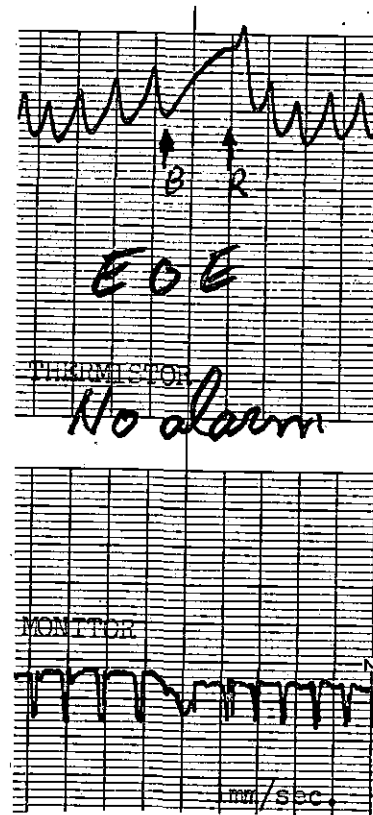
(a)



(b)



(c)



(d)

activated. It should be mentioned that the monitor time threshold was set at 7 seconds and no computer was used for detecting U.A.O.

In the next experiment, a minocomputer was used for detection of U.A.O. The results are as follows. First, normal respiration was recorded; it was noticed that occasionally, because of external noise, the phase of the monitor waveform changed 180° , as shown in Figure 10. Denote the resultant phases "phase a" and "phase b". Also, note the presence of heart beats in the monitor waveform. Then the upper airway was occluded several times. The results showed that when the waveform was in "phase a" and there was no phase shift during the occlusion, the computer detected U.A.O. and triggered the computer buzzer (Figure 11a). When the waveform was in "phase b" and there was no shift to "phase a", the computer did not detect the occlusion (Figure 11b). When there was a shift from "phase a" to "phase b", two out of five applications of U.A.O. were detected by the computer, while only one U.A.O., was detected by the monitor alarm (Figure 12). When there was a shift from "phase b" to "phase a", out of four applications of U.A.O., one was detected by the computer and one was detected by the monitor alarm (Figure 13). Note that in all figures (Figures 8-13), heart beats are recorded.

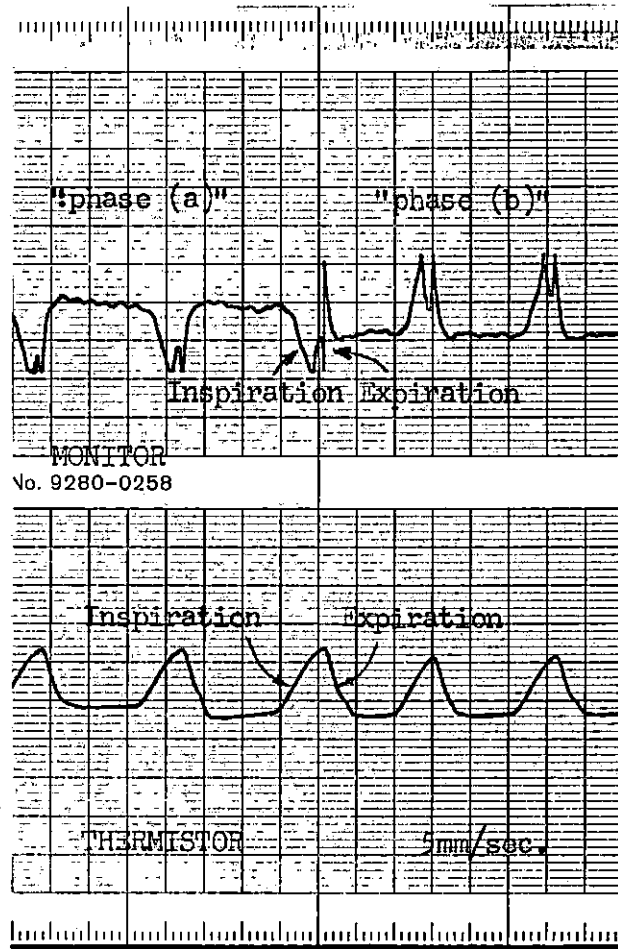


Figure 10. Occasional change in the monitor waveform phase

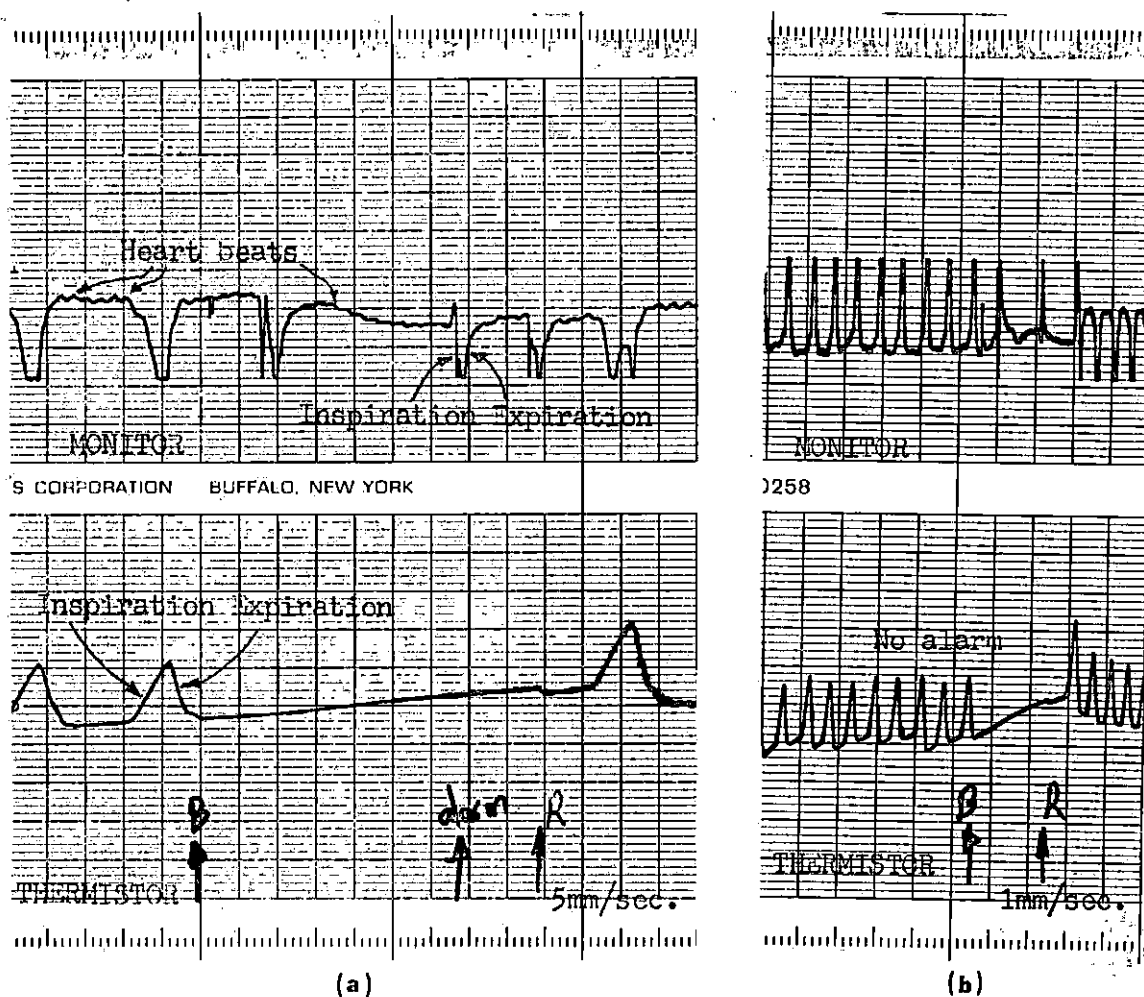


Figure 11. Detection of U.A.O. when no phase shift is present: (a) waveform in "phase a", (b) waveform in "phase b". The computer detects U.A.O. if the monitor waveform is in "phase a" and the phase does not shift. Blockage was applied at point B and removed at point R

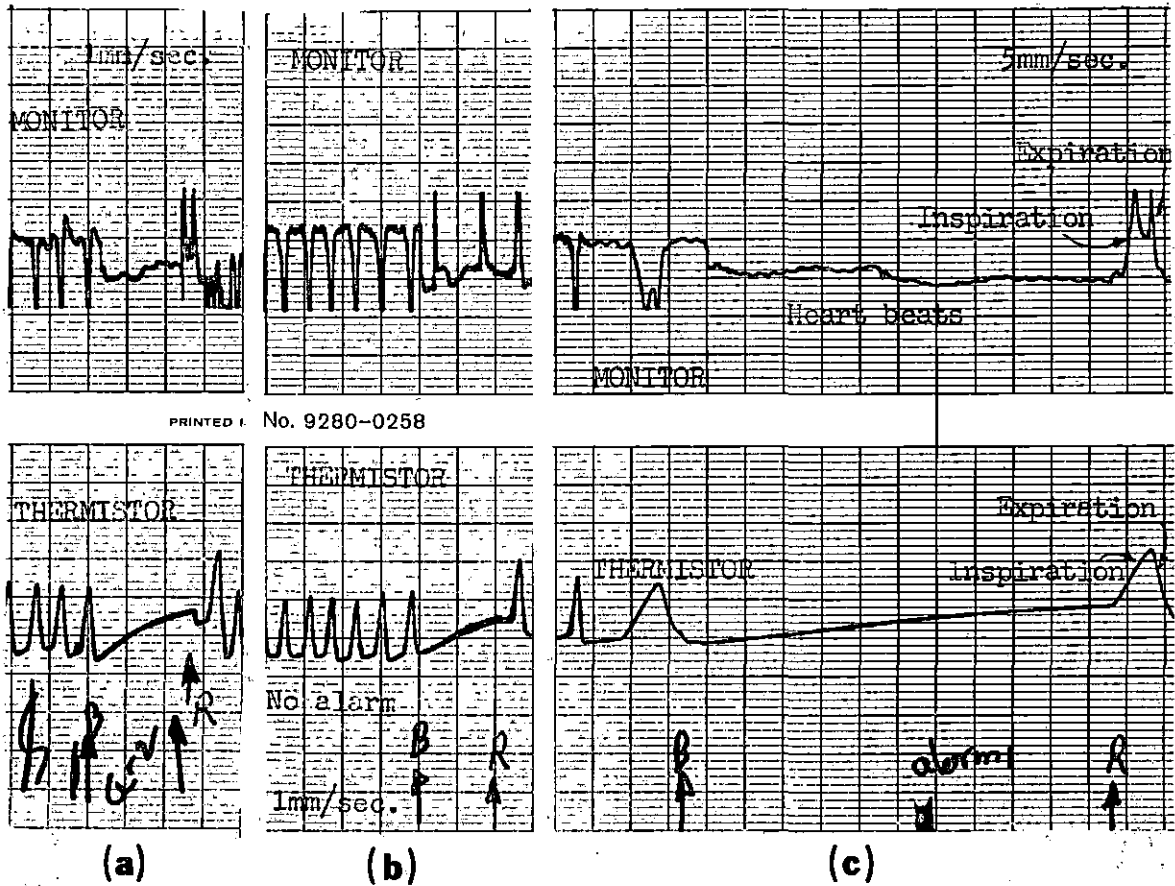


Figure 12. Detection of U.A.O. when a shift from "phase a" to "phase b" occurs. Case (a) was detected by the computer. Case (b) was not detected. Case (c) was detected by the monitor alarm. Blockage was applied at point B and removed at point R.

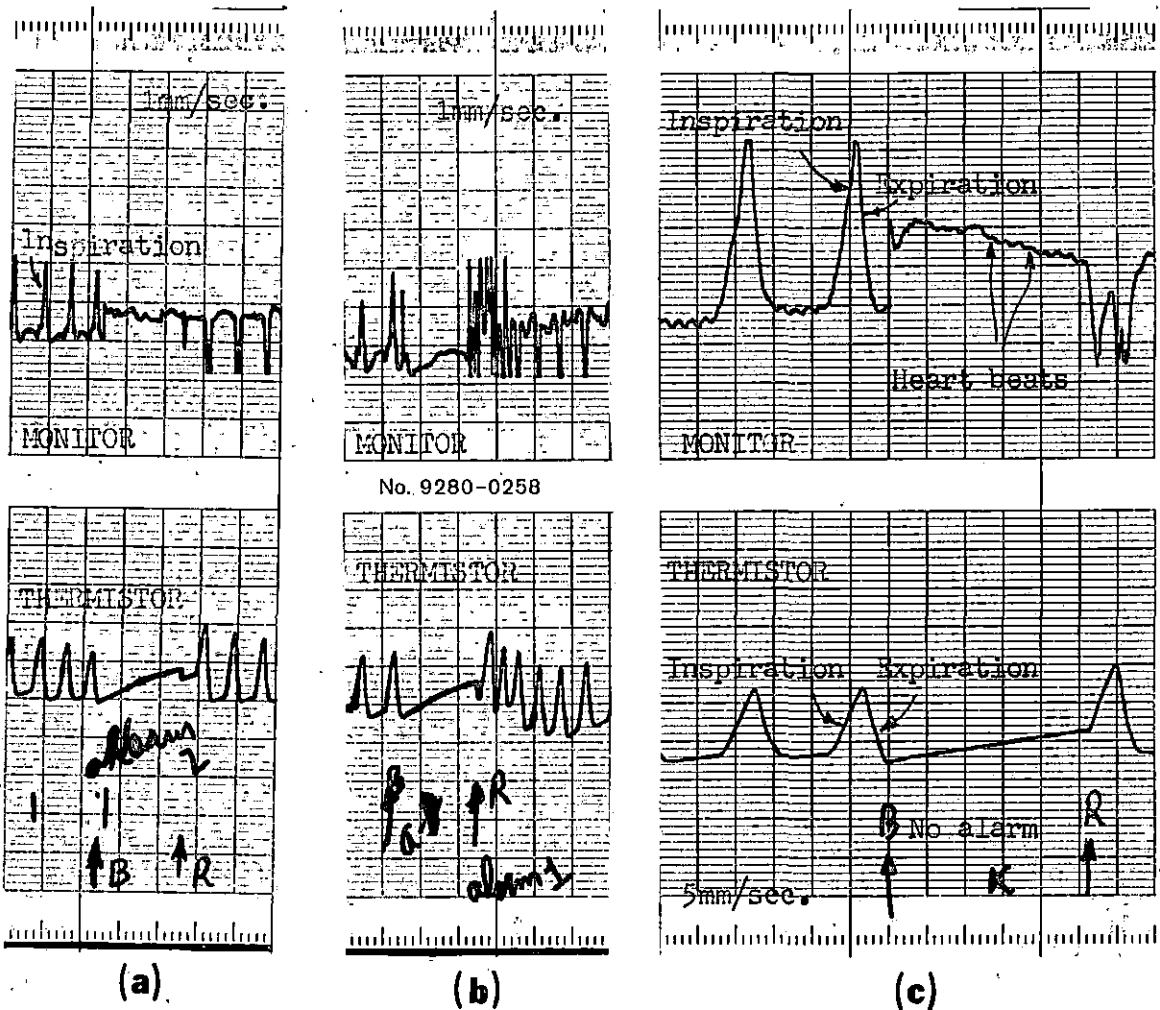


Figure 13. Detection of U.A.O. when a shift from "phase b" to "phase a" is present. Case (a) was detected by the computer. Case (b) was detected by the monitor alarm. Case (c) was not detected. Blockage was applied at point B and removed at point R.

DISCUSSION

Implications of Results

The performance of the ultrasonic apnea monitor indicates that under certain conditions it detects respiration motions very well and reproduces the respiration patterns acceptably. In all experiments on dogs there was not a single incident of false alarm, even in cases where respiration was shallow or the dog was lain in a position such that the abdomen wall displacement was very small.

Figure 8(a) supports such a statement.

The monitor is sensitive enough to pick up the heart beats, as shown in Figures 8 through 13, but the alarm section ignores such motions. Figure 12(c) and 12(b) are good examples of such cases that during a period of apnea even though the heart beat was detected and recorded, the monitor alarm was activated. It is the presence of respiration, not the heart beat, which turns off the activation of the alarm.

The faithfulness of reproduction of the respiration pattern is demonstrated in Figures 8-13. The slight difference between the widths of the monitor waveform and the thermistor pattern is due to the slow response of the thermistor. The different waveforms (double peaking waveform) of Figure 7 (and others) are also due to the small wavelength of the ultrasonic signal. As mentioned in the

previous chapter, if the abdomen wall displacement is larger than half the ultrasonic wavelength, the phase angle difference exceeds 2π radians and repeats itself; thus, a double peaking waveform is generated. This problem was solved by adding a pair of frequency dividers to the system. The frequency dividers increase the apparent wavelength and thus eliminate the problem. (There was no frequency divider in the circuit when the recording shown in Figure 7 was made.)

In relation to faithfulness of the reproduction of the respiration waveforms, one may point to Figures 8(a) and 8(b) as counter examples. Figure 8(a) shows that when the transducers are facing the side of the dog respiration is detected but the waveforms are not similar to the waveforms of the thermistor.

Figure 8(b) shows that if the transducers are facing the chest of the dog, the monitor can not detect respiration, mainly because of the shape of the thoracic bone of the dog which reflects the transmitted signals to a direction different from the position of the receiver transducer.

Thus we come to the conclusion as suggested by Figure 8(c) that the best position for monitoring respiration of a dog, using this monitor, is the supine position while the transducers are facing that part of abdomen immediately

below the thorax. It is this position which is used for detection of upper airway occlusion.

Upper airway occlusion can be detected under certain conditions. The monitor, without the help of a computer, can detect U.A.O. if the occlusion occurs at any phase of respiration except at the end of expiration and the monitor alarm is set for a 10 second or shorter apnoeic attack. Figures 7(b) and 7(c) and Figure 9 can be used to show the reasons, in this effect. Figure 7(b) shows that when the airway is occluded at the end of inspiration no immediate respiratory effort is made by the dog, a condition which triggers the alarm. The lack of respiration is due to the stimulation of inflation receptors in the lungs which inhibit respiration.

Figure 7(c) shows a recording of respiration when the upper airway is blocked at the end of expiration. Respiratory efforts are still present, because respiratory center is not inhibited by the inflation receptors in the lungs. Yet the pattern is considerably different from that of Figure 7(b). It is this case which is not detected by most motion detecting monitors.

Figure 9 shows recordings of U.A.O. when applied at different phases of respiration. In Figure 9(a) the airway is blocked in the middle of expiration. The monitor alarm was triggered, but was deactivated when the next effort for

respiration was made. The case of Figure 9(b) had the same kind of response. In Figure 9(c) the alarm was triggered and was not deactivated until the blockage was removed. In the case of Figure 9(d), the alarm remained silent during U.A.O.

For the U.A.O. applied at the end of expiration to be detected, a minicomputer should be used, but first the specific changes in the patterns of respiration during U.A.O. should be studied to permit an appropriate computer program to be written. Figure 14 shows the typical changes in the respiration waveform, when U.A.O. is applied.

It should be noted that the pattern of respiration during U.A.O. changes in respect to both width and rate. When a respiratory effort is initiated, the chest begins to expand, but the upper airway is blocked and the chest cannot expand easily. Since the inflation receptors are not stimulated to inhibit respiration, the chest continues to expand, and thus the width of the respiratory pattern increases. Tests have shown that the increase is at least 50% of the original width.

For the same reasons, the period of the waveform increases. But the second respiratory effort after application of U.A.O. is initiated sooner and stronger compared to normal respiratory efforts. Therefore, only the first cycle after application of U.A.O. undergo significant changes.

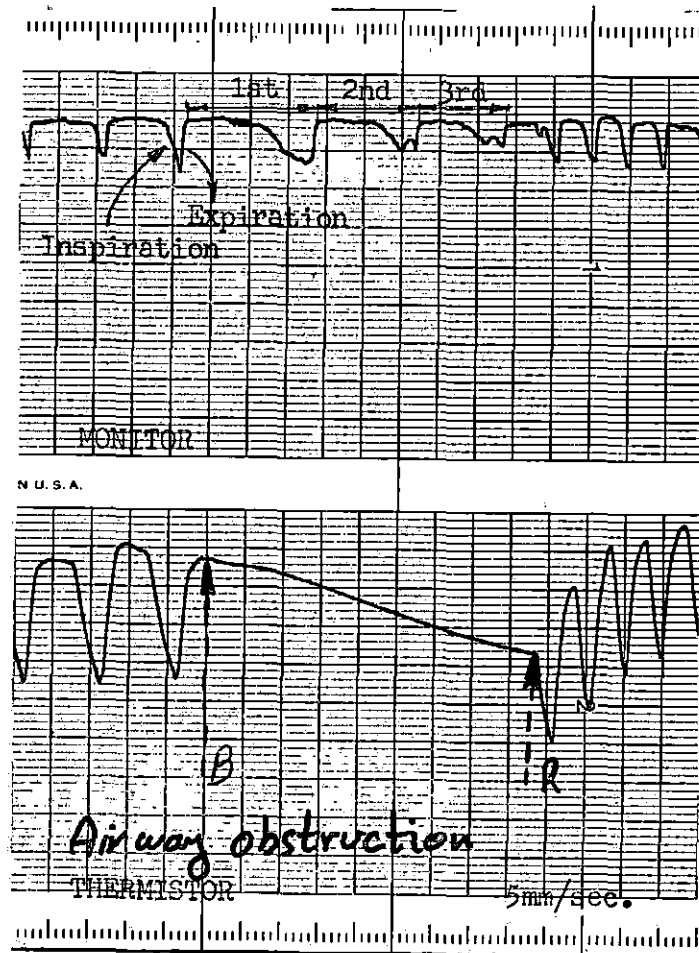


Figure 14. Changes in the respiration pattern when upper airway is occluded at the end of expiration. The width increases and the rate decreases. Note that the 2nd cycle of respiration after blockage was applied is smaller in period than the 1st cycle immediately after U.A.O. Blockage was applied at point B and removed at point R

These changes in the respiratory pattern cannot be mistaken for changes due to gasps or sighs, because in gasps and sighs only the amplitude of the waveform increase, not the width.

The minicomputer has shown to be effective in detecting U.A.O. However, there are cases which may not be detected by the computer. Phase shift is involved in such situations. The phase shift is believed to be due to external electrical noise which makes the monostable multivibrator output (described on page 23) to go high on the edge of the second cycle of the input signal, and thus changing the phase angle. As shown in Figure 10, when phase angle changes, a different threshold is required for detection of the onset and the end of each cycle of respiration pattern, and thus the computer program used is unable to detect U.A.O.

However, in cases where phase shift is present, U.A.O. may still be detected, e.g., Figures 12(a), 12(c), 13(a) and 13(b). Among these cases, 12(c) and 13(b) were detected by the monitor, and the rest were detected by the computer. The computer detected these cases because the threshold needed for the detection of the onset and the end of each cycle of respiration had changed very little, a condition in which the computer could distinguish between different cycles of respiration from each other (see Figures 12(c) and

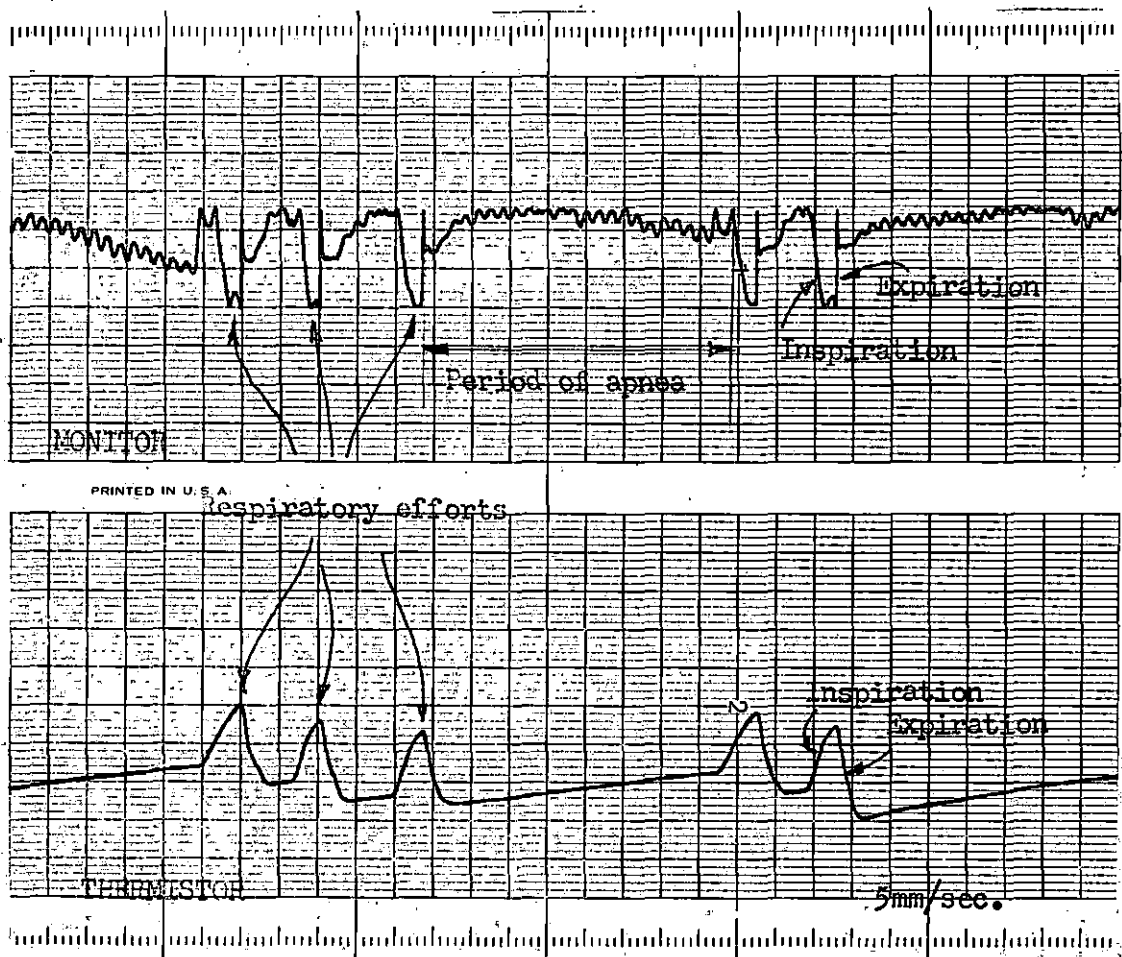


Figure 15. Biot breathing pattern. In this pattern, several respiratory efforts are made followed by a period of apnea.

13(a)).

Figure 12(b) shows a case in which, even though the threshold change was small, U.A.O. was not detected, because what is considered as the width of the signal in "phase a", is considered the period of relaxation of respiration in "phase b". Thus, when a phase shift from "phase a" to "phase b" occurs, since there is an apparent decrease in the signal width, the computer does not consider such a change a blockage. The computer could be programmed to detect an increase in the period of relaxation, as a blockage, but then the Biot breathing pattern cannot be detected. Figure 15 shows a sample of Biot breathing pattern which was recorded during one of the tests. In this pattern there is a burst of respiration and then a period of apnea. The computer has been programmed to detect Biot breathing pattern and disregard it. In Figure 12(b), the case appears to the computer to be a Biot breathing pattern and is disregarded.

Conclusion and Suggestions for Improvements

Apnoeic attack, the absence of breathing for a period of time, is usually seen in premature infants during the first days or even during the first hours after birth. It is a condition which in most cases results in the death of

the infant or may cause hypoxic brain damage among survivors. In orders to prevent such disasters, apnea monitors are instituted to detect apnoeic attacks and to provide the physician with essential information for treating the patient.

Among different monitors, invasive techniques are the most reliable methods, but they pose additional threats, such as infection or hemorrhage, to the infant. Therefore, contactless monitors are more desirable. The ultrasonic apnea monitor described in this thesis carries neither the threat of the invasive methods nor many of the disadvantages of other contactless techniques.

The ultrasonic apnea monitor utilizes a pair of air-pressure transducers. One of the transducers sends pressure waves to the infant's chest or abdomen. The reflected waves are picked up by the second transducer. The monitor continuously monitors the phase difference between the transmitted and the reflected waves. Whenever the phase difference remained constant, a condition indicating no respiration, the monitor activates the alarm. In addition to this monitor, a minicomputer is utilized to detect the U.A.O. which are not detected by the monitor.

As indicated in the discussion of the results, the ultrasonic apnea monitor is a reliable device which can detect respiration, reproduce the respiration pattern.

faithfully, and trigger an alarm whenever apnoeic attack occurs. It functions best when the transducers are facing the area just under the thorax when small dogs are used as the test animal.

The monitor can detect U.A.O. if U.A.O. occurs at any phase of respiration except at the end of expiration, and the monitor alarm is set for a 10 second or shorter apnoeic attack. The computer program can detect U.A.O. occurred at the end of expiration if certain conditions are met. These are: 1) the waveforms should be in "phase a"; and 2) the phase should not shift. The first condition can be met simply by adjusting the distance between the transducers and the abdomen. The second condition can be met by either removing the frequency dividers, which causes the double peaking effect to reappear, or by filtering out the external noise.

A better suggestion for eliminating the phase shift phenomenon is to replace the transducers with lower frequency operating transducers, e.g., 25 KHz, which increases the wavelength, and thus eliminating the phase shift problem. It should also be noted that decreasing the frequency may cause some hearing problems for the patient when the upper limit of hearing sensitivity is approached.

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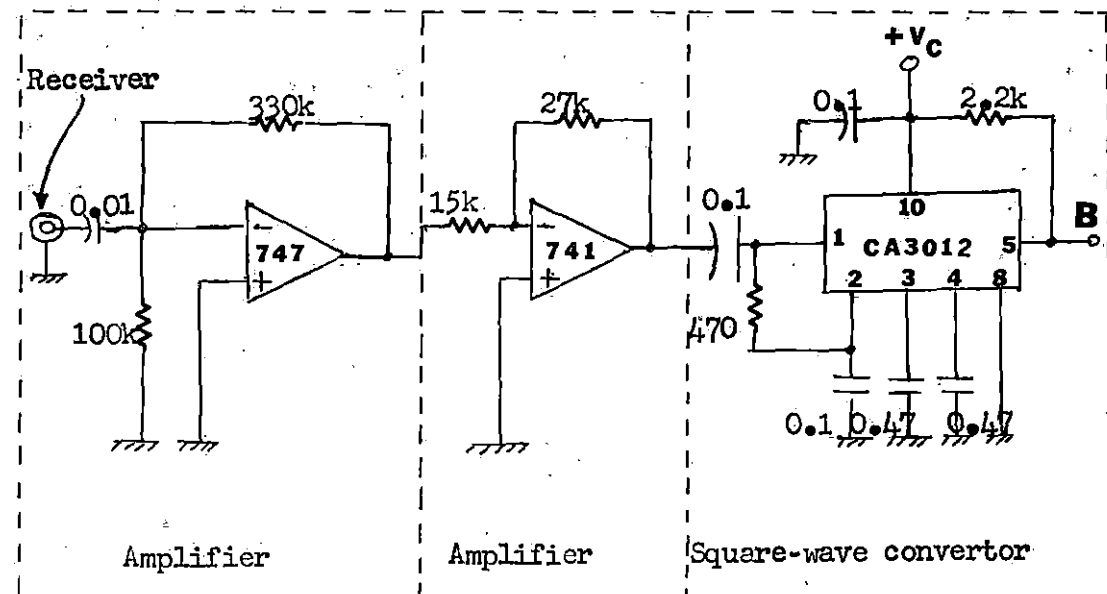
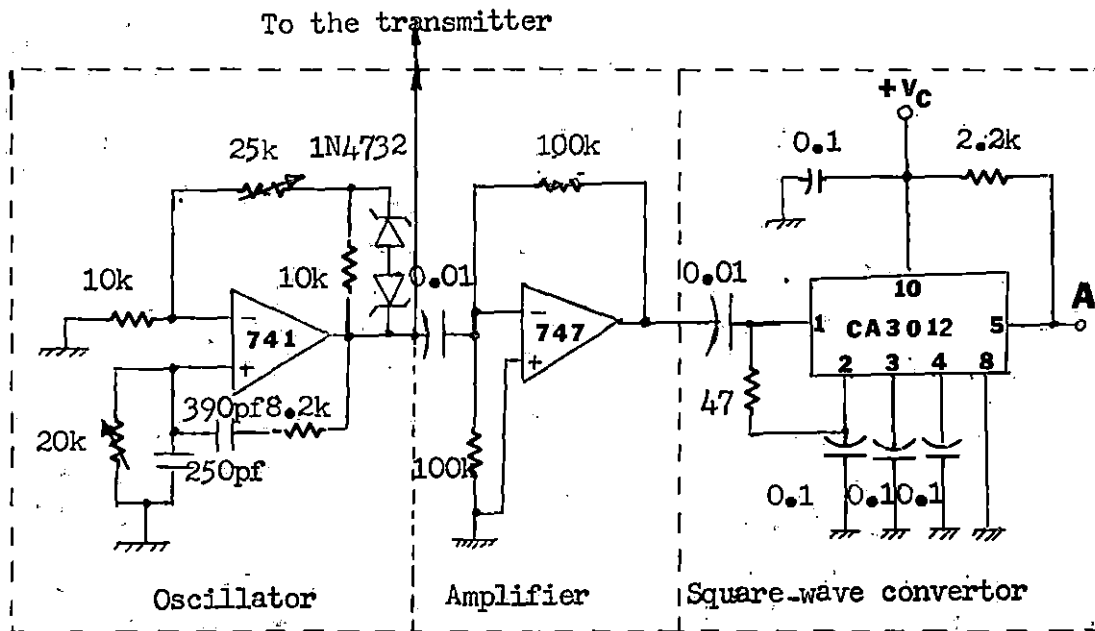
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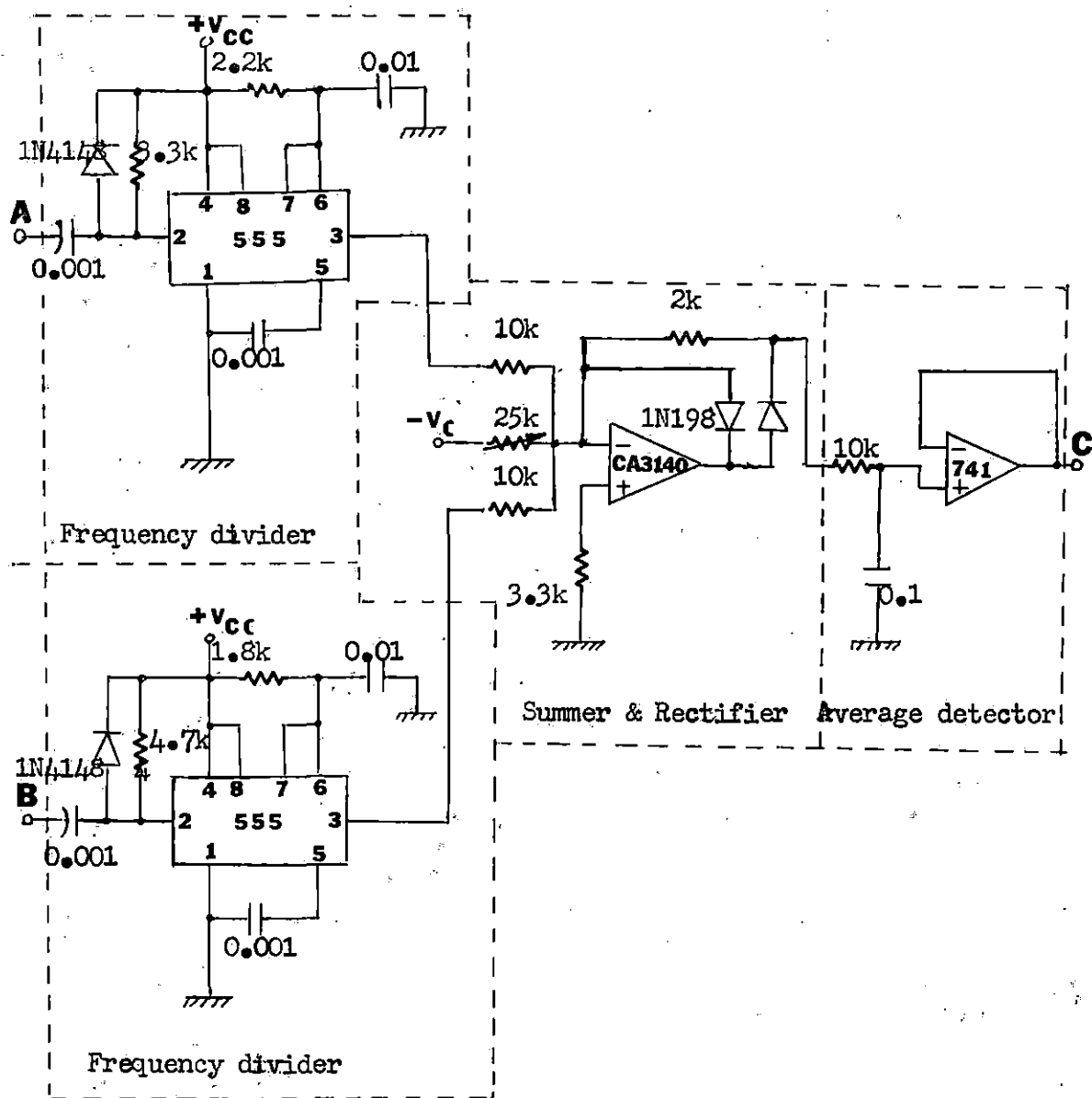
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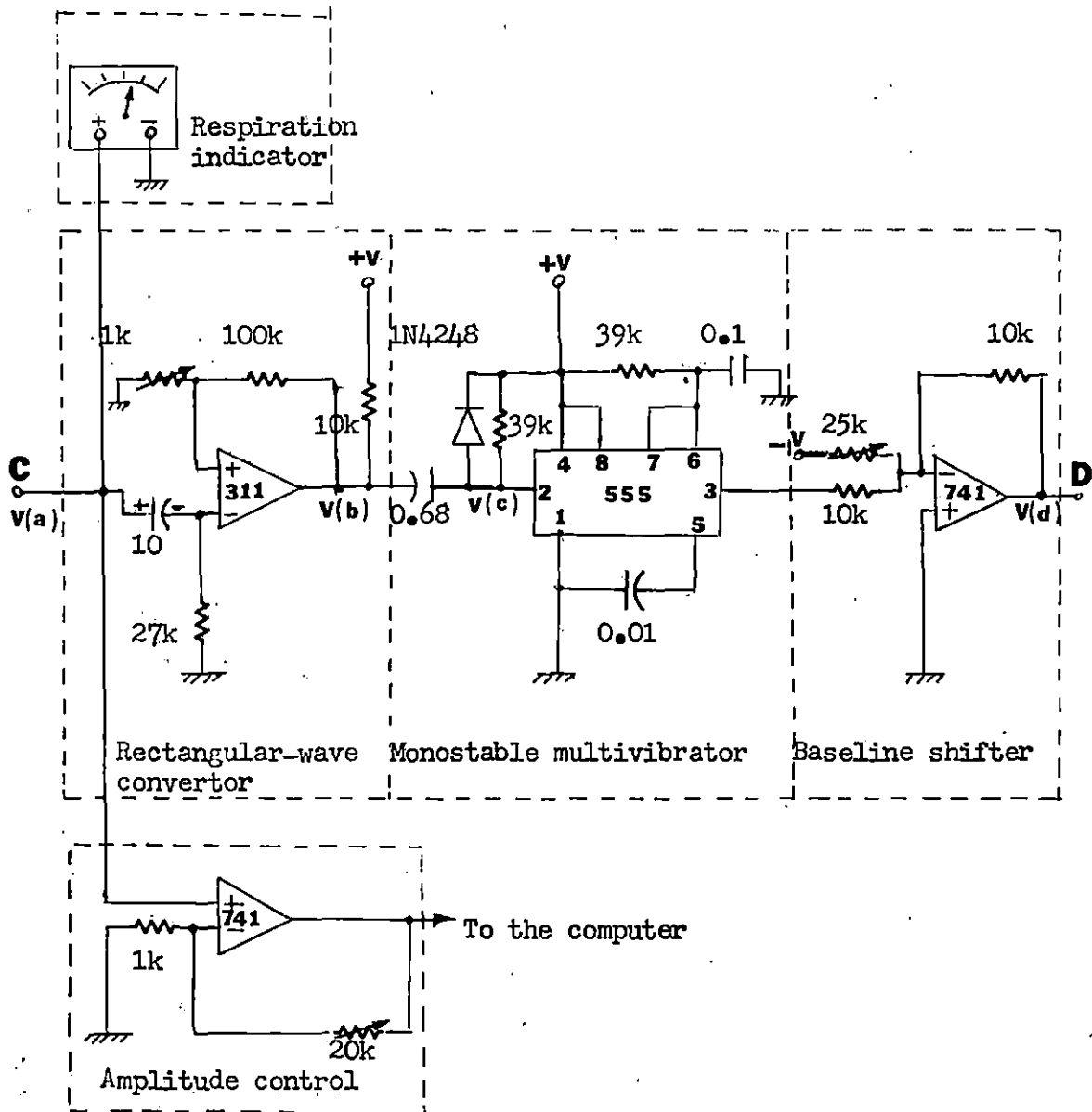
APPENDIX A: ELECTRONIC CIRCUITRY



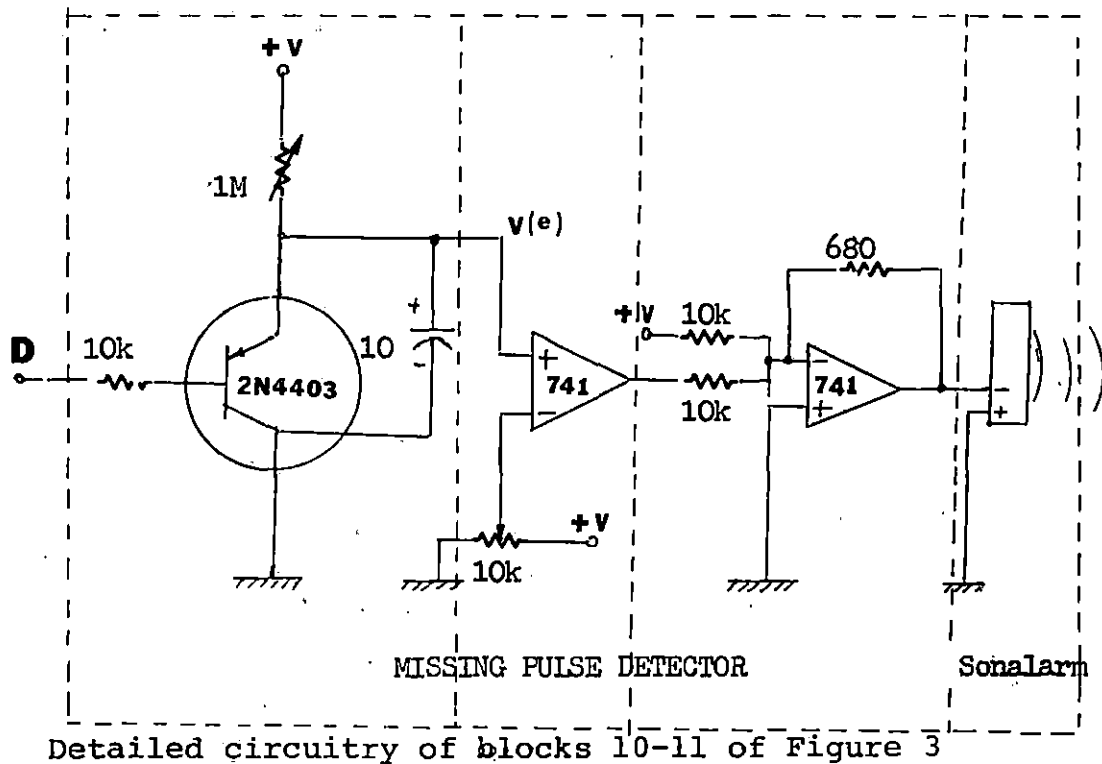
Detailed circuitry of blocks 1-3 of Figure (all resistors are OHMS and all capacitors in microFARADS unless otherwise mentioned)



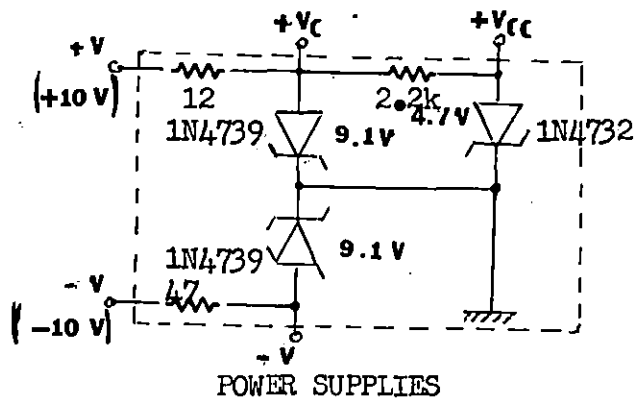
Detailed circuitry of blocks 4-7 of Figure 3



Detailed circuitry of blocks 8-9 of Figure 3

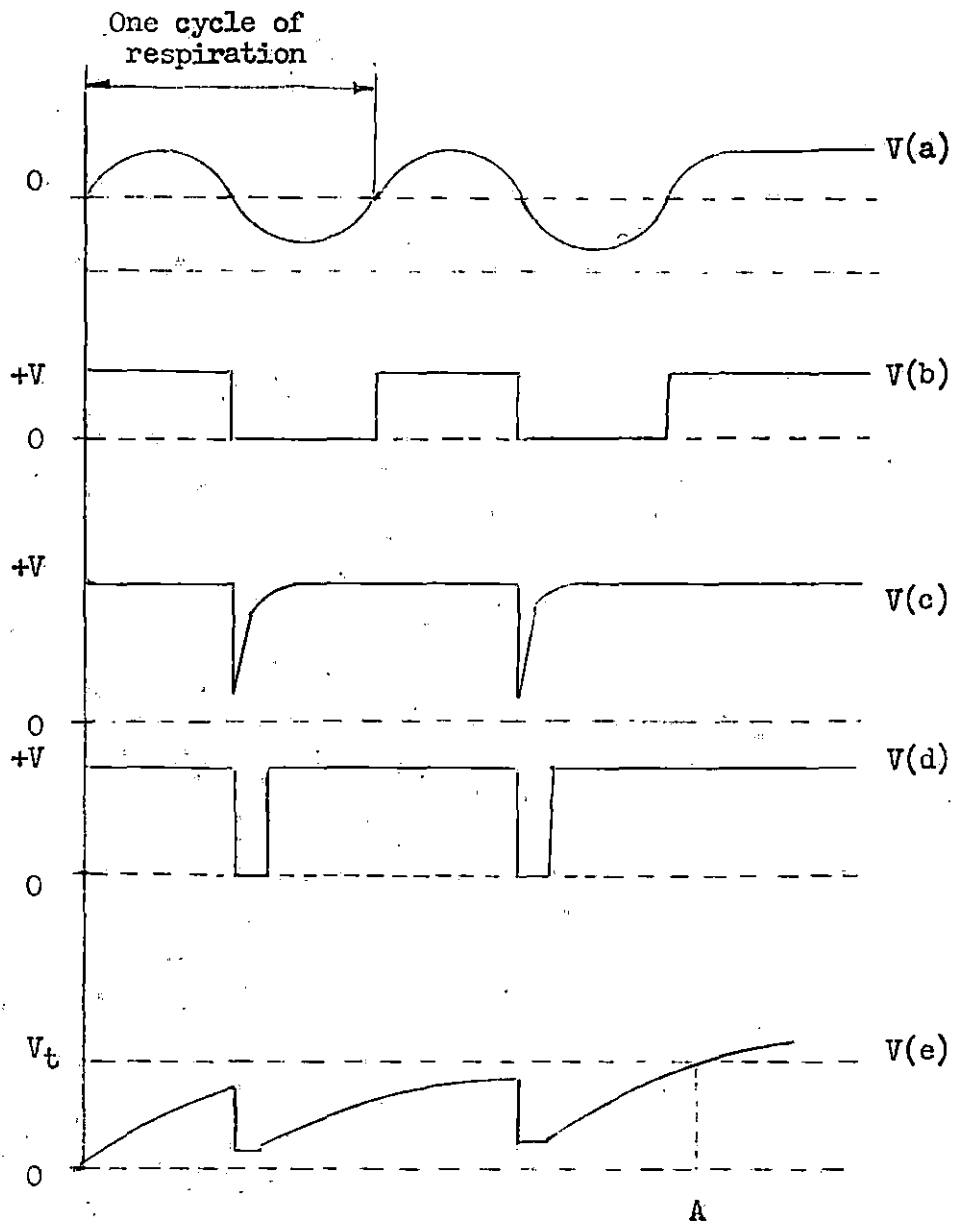


Detailed circuitry of blocks 10-11 of Figure 3

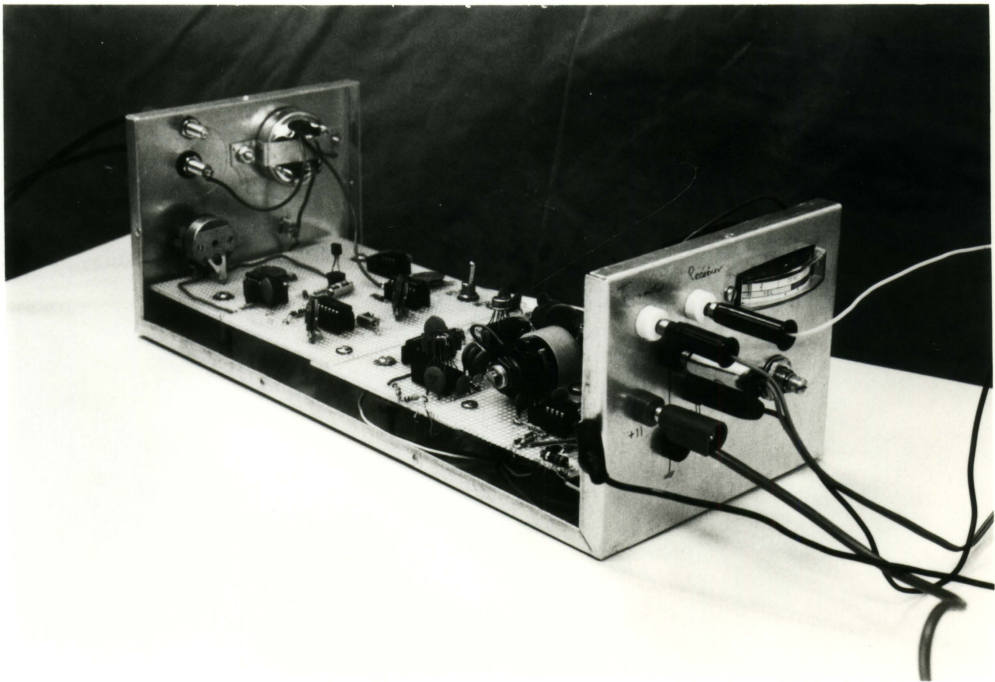


POWER SUPPLIES

+V and -V are used in the alarm section, +V_C and -V_C are used in the rest of the circuit



Waveforms of signals at several points of the alarm circuitry. The alarm is activated at point A when V(e) exceeds the threshold voltage (V_t)



The ultrasonic apnea monitor.

APPENDIX B: COMPUTER PROGRAMS

00020	0020	*20	000	/ SUBROUTINE TO WAIT 10 MSEC.
00021	0000	WAIT,	CLA	
00022	7200		NOP	
00023	7000		ISZ W	
00024	2125		JMP .-2	
00025	5022		TAD MW	
00026	1126		DCA W	
00027	3125		ISZ Y	
00027	2127		JMP .-6	
00030	5022		TAD MY	
00031	1130		DCA Y	
00032	3127		JMP I WAIT	/ SUBROUTINE TO PRINT THE CONTENTS
00033	5420	TYPE,	000	/ OF THE ACCUMULATOR.
00034	0000		TSF	
00035	6041		JMP .-1	
00036	5035		TLS	
00037	6046		JMP I TYPE	/ SUBROUTINE TO ACTIVATE THE BUZZER.
00040	5434	BELL,	000	
00041	0000		CLA	
00042	7200		TAD K207	
00043	1123		JMS TYPE	
00044	4034		6501	
00045	6501		JMP .-4	
00046	5042		HLT	
00047	7402	START,	CLA	/ THIS PORTION INITIALIZES THE
00050	7200		DCA MAG	/ PROGRAM.
00051	3074		DCA T	/ T IS THE WIDTH OR THE PERIOD.
00052	3075		DCA QUOT1	/ QUOT1 IS THE AVG. OF THE WIDTHS.
00053	3110		DCA QUOT2	/ QUOT2 IS THE AVG. OF THE PERIODS.
00054	3111		DCA SUMA	
00055	3106		DCA SUMT	
00056	3107		TAD MTS	/ MTS & TS ARE THE LOCATIONS THE
00057	1077		DCA TS	/ PERIODS ARE STORED.
00060	3076		TAD MA	/ MA & AS ARE THE LOCATIONSTHE WIDTHS
00061	1101		DCA AS	/ ARE STORED.
00062	3100		TAD MCOUNT	
00063	1103		DCA COUNT	
00064	3102		TAD MB	/ BS IS THE LOCATION OF THE 2nd WIDTH.
00065	1112		DCA BS	
00066	3113		TAD MC	/ C IS THE LOCATION OF THE 2nd PERIOD.
00067	1115		DCA C	
00070	3114		TAD MCODE	
00071	1122		DCA CODE	
00072	3121		JMP BEGIN	
00073	5777			
00074	0000	MAG,0		
00075	0000	T,0		
00076	0170	TS,170		/ STARTING LOCATION OF THE PERIODS.
00077	0170	MTS,170		
00100	0160	AS,160		/ STARTING LOCATION OF THE WIDTHS.

00101	0160	MA,160
00102	7772	COUNT,-6
00103	7772	MCOUNT,-6
00104	7773	M5,-5
00105	0005	M2,5
00106	0000	SUMA,0
00107	0000	SUMT,0
00110	0000	QUOT1,0
00111	0000	QUOT2,0
00112	0161	MB,161
00113	0161	BS,161
00114	0171	C,171
00115	0171	MC,171
00116	7777	M1,-1
00117	0165	M165,165
00120	0175	M175,175
00121	0140	CODE,140
00122	0140	MCODE,140
00123	0207	K207,207
00124	7773	NUM5,-5
		DECIMAL
00125	7406	W,-250
00126	7406	MW,-250
00127	7770	Y,-8
00130	7770	MY,-8
		OCTAL
	0140	*140
00140	0301	
00141	0311	
00142	0322	
00143	0327	
00144	0301	
00145	0331	
00146	0240	
00147	0302	
00150	0314	
00151	0317	
00152	0303	
00153	0313	
00154	0305	
00155	0304	
00156	0215	
00157	0212	
00160	0000	

/ AVG. OF THE WIDTHS.
 / AVG. OF THE PERIODS.
 / LOCATION OF THE 2ND WIDTH.
 / LOCATION OF THE 2ND PERIOD.
 / LOCATION OF THE 6TH WIDTH.
 / LOCATION OF THE 6TH PERIOD.

301
 311
 322
 327
 301
 331
 240
 302
 314
 317
 303
 313
 305
 304
 215
 212
 000

00200	7200	BEGIN,	CLA	/ BEGIN FINDS THE ONSET OF INSPIRATION.
00201	6560		6560	
00202	3074		DCA MAG	
00203	6561		6561	
00204	1074		TAD MAG	
00205	7510		SPA	
00206	5200		JMP BEGIN	
00207	7200	SURFACE,	CLA	/ IN THIS SECTION THE WIDTH IS MEASURED.
00210	4020		JMS WAIT	
00211	2075		ISZ T	
00212	6560		6560	
00213	3074		DCA MAG	
00214	6561		6561	
00215	1074		TAD MAG	
00216	7500		SMA	
00217	5207		JMP SURFACE	
00220	7200		CLA	
00221	1075		TAD T	
00222	2100		ISZ AS	
00223	3500		DCA I AS	
00224	1104		TAD M5	
00225	3124		DCA NUM5	
00226	4020	PERIOD,	JMS WAIT	/ IN THIS SECTION THE PERIOD IS MEASURED
00227	2075		ISZ T	
00230	6560		6560	
00231	3074		DCA MAG	
00232	6561		6561	
00233	1074		TAD MAG	
00234	7510		SPA	
00235	5226		JMP PERIOD	
00236	7200		CLA	
00237	1075		TAD T	
00240	2076		ISZ TS	
00241	3476		DCA I TS	
00242	3075		DCA T	
00243	2102		ISZ COUNT	
00244	5207		JMP SURFACE	
00245	7200		CLA	
00246	1104		TAD M5	
00247	3102		DCA COUNT	
00250	1077		TAD MTS	
00251	3076		DCA TS	
00252	1101		TAD MA	
00253	3100		DCA AS	
00254	2100	ADD,	ISZ AS	/ THIS PORTION FINDS THE SUM OF THE 1st
00255	2076		ISZ TS	/ FIVE WIDTHS AND THE SUM OF THE 1st FIVE
00256	1500		TAD I AS	/ PERIODS.
00257	1106		TAD SUMA	
00260	3106		DCA SUMA	

00261	1476	TAD ITS	
00262	1107	TAD SUMT	
00263	3107	DCA SUMT	
00264	2102	ISZ COUNT	
00265	5254	JMP ADD	
00266	1106	TAD SUMA	
00267	7041	CIA	
00270	1105	DVIDE1, TAD MQ	/ THIS PORTION FINDS THE AVG. OF
00271	2110	ISZ QUOT1	/ THE WIDTHS.
00272	7510	SPA	
00273	5270	JMP DVIDE1	
00274	7200	CIA	
00275	1107	TAD SUMT	
00276	7041	CIA	
00277	1105	DVIDE2, TAD MQ	/ THIS PORTION FINDS THE AVG. OF
00300	2111	ISZ QUOT2	/ THE PERIODS.
00301	7510	SPA	
00302	5277	JMP DVIDE2	
00303	7300	CIA CIL	/ THIS PORTION CHECKS THE WIDTH FOR
00304	1110	TAD QUOT1	/ 50% INCREASE. IF YES CHECKS THE PERIOD.
00305	7010	RAR	
00306	1110	TAD QUOT1	
00307	7041	CIA	
00310	2100	ISZ AS	
00311	1500	TAD I AS	
00312	7500	SMA	/
00313	5374	JMP TEST	
00314	1110	TAD QUOT1	/ THIS PORTION CHECKS THE WIDTH
00315	7510	SPA	/ FOR 50% DECREASE. IF YES IGNORES THE
00316	5351	JMP CONT	/ CYCLE.
00317	7300	CIA CIL	
00320	1111	TAD QUOT2	/ THIS PORTION CHECKS THE PERIOD
00321	7010	RAR	/ FOR 50% INCREASE. IF YES IGNORES
00322	1111	TAD QUOT2	/ THIS CYCLE.
00323	7041	CIA	
00324	2076	ISZ TS	
00325	1476	TAD I TS	
00326	7500	SMA	
00327	5351	JMP CONT	
00330	7200	INITIAL, CLA	/ THIS PORTION INITIALIZES THE PROGRAM
00331	1104	TAD M5	/ FOR GOING THROUGH THE LOOP, AND SHIFTS
00332	3102	DCA COUNT	/ THE STORED DATA 1 BYTE.
00333	1101	TAD MA	
00334	3100	DCA AS	
00335	1077	TAD MTS	
00336	3076	DCA TS	
00337	2076	ISZ TS	
00340	2100	ISZ AS	

00341	2114	ISZ C	
00342	2113	ISZ BS	
00343	1513	TAD I BS	
00344	3500	DCA IAS	
00345	1514	TAD IC	
00346	3476	DCA ITS	
00347	2102	ISZ COUNT	
00350	5337	JMP .-11	
00351	7200	CONT, CLA	/ THIS PORTION INITIALIZES THE PROGRAM
00352	1116	TAD M1	/ WHEN THE LAST CYCLE OF RES PIRATION
00353	3102	DCA COUNT	/ IS IGNORED.
00354	1115	TAD MC	
00355	3114	DCA C	
00356	1112	TAD MB	
00357	3113	DCA BS	
00360	1117	TAD M165	
00361	3100	DCA AS	
00362	1120	TAD M175	
00363	3076	DCA TS	
00364	3110	DCA QUOT1	
00365	3111	DCA QUOT2	
00366	3106	DCA SUMA	
00367	3107	DCA SUMT	
00370	6502	6502	
00371	5207	JMP SURFACE	
00372	7402	HLT	
00373	5050	JMP START	
00374	7300	TEST, CLA CLL	/ THIS PORTION CHECKS FOR 50%
00375	1111	TAD QUOT2	/ INCREASE IN THE PERIOD.
00376	7010	RAR	
00377	1111	TAD QUOT2	
00400	7041	CIA	
00401	2076	ISZ TS	
00402	1476	TAD I TS	
00403	7500	SMA	
00404	5206	JMP PRINT	
00405	5777	JMP CONT	
00406	7200	PRINT, CLA	/ THIS PORTION PRINTS "AIRWAY BLOCKED".
00407	6046	TLS	
00410	7200	CIA	
00411	1521	TAD I CODE	
00412	7450	SNA	
00413	4041	JMS BELL	
00414	4043	JMS TYPE	
00415	2121	ISZ CODE	
00416	5210	JMP .-6	
00577	0350		
	\$		
00177	0200		

ADD	0254
AS	0100
BEGIN	0200
BELL	0041
BS	0113
C	0114
CODE	0121
COT	0351
COUNT	0102
DVIDE1	0270
DVIDE2	0277
INITIA	0330
K207	0123
MA	0101
MAG	0074
MB	0112
MC	0115
MCODE	0122
MCOUNT	0103
M2	0105
MTS	0077
MW	0126
MY	0130
M1	0116
M165	0117
M175	0120
M5	0104
NUM5	0124
PERIOD	0226
PRINT	0406
QUOT1	0110
QUOT2	0111
START	0050
SUMA	0106
SUMT	0107
SURFAC	0207
T	0075
TEST	0374
TS	0076
TYPE	0034
W	0125
WAIT	0020
Y	0127

In periphery to the computer 2 channels of the A/D convertor and two sense switches are used. In channel 0 of the A/D convertor, the respiration waveform from the monitor is plugged. The potentiometer connected to Channel 1 is used for setting the threshold. The settings are:

	Potentiometer Setting											
	1.92	2	2.1	2.2	2.3	2.4	3.38	3.6	3.8	4.04	4.4	4.5
Volts	-7.52	-6.75	-6.51	-6.38	-6.15	-5.42	-4	-3.5	-3	-2.5	-2	-1.5

The first sense switch, if down, discontinues the computer buzzer. The second switch, if down, halts the execution of the program. Then, if the continue switch is pressed, the program would be run from the start.

In order to run the program, the same switches should be up and the switch to the left of the register switches on the minicomputer should be down. The program starts at location 50.